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Impact Biomechanics of the Pelvis and Lower Limbs in Occupants Involved in an Impact Aircraft Accident

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ABSTRACT

Impact biomechanics of the pelvis and lower limbs in occupants involved in an aircraft accident have been investigated using a variety of techniques. These techniques have been used to:

1) Explore whether the position adopted by the occupant of the plane at the time of impact had implications for the pelvic and lower limb injuries sustained.

2) Test and assess the relevance of hypothesised injury mechanisms for the pelvis and lower limbs, described in the automobile industry to that of an impact aircraft accident.

Clinical data has been derived form a cohort of accident victims on board Boeing 737-400, G-OBME, when it crashed on the M1 motorway on the 8 January 1989. Experimental impact testing has been carried out using anthropomorphic test devices and a deceleration sled test facility. Further investigation of the impact biomechanics has utilised new techniques of impact occupant modelling with the aid of computer simulations.

The results have indicated that in areas of the aircraft

where seating and restraint mechanisms remained intact and fuselage disruption was minimal, severe lower limb and pelvic injuries were sustained by the occupants. These injuries may have been sustained in the absence of significant secondary impacts of the lower limbs with the seat in front.

Further experiments have indicated that the position adopted by the occupants, and in particular the placement of the lower limbs on the floor can affect the trajectories of the limbs in their flail behaviour. In addition it is apparent that the knee-femur-pelvis mechanism of lower limb injury recognised by the automobile industry may not have been an important mechanism in this aviation situation.

These findings have implications for the design of occupant safety systems if pelvic and lower limb injuries are to be reduced in future aircraft accidents.

CANDIDATE'S DECLARATION

The work contained within this thesis by its nature has involved collaboration with many individuals and bodies. Review of injuries to the occupants has on the whole been carried out by myself. The experimental planning and conduct has been undertaken and supervised by myself, although technical advice and assistance were required. The analysis, interpretation and conclusions drawn from this study are entirely my own.

Acknowledgements to those individuals and bodies who aided me in the production of this thesis are made overleaf. However special assistance was received from Andrew Belyavin (Statistics Department, IAM Farnborough) who performed the analysis of variance on the sled test experiments and Raf Haidar (HW Structures, Lemington Spa) who carried out the computer simulations.

All the work was carried out during my appointment as Research Registrar to the NLDB Study Group, University Hospital Nottingham, during the period 1989 to 1990.

ACKNOWLEDGEMENTS

This work would not have been possible without the help and advice of many people. I must first record my thanks to Professor W Angus Wallace, who had the courage to appoint me as a Research Registrar, when my experience in research was limited. In my time as Research Registrar he has been instrumental in encouraging me to carry out the work for this thesis and supervising its completion.

Great thanks must also be extended to the Institute of Aviation Medicine, Farnborough, who have allowed me to use the facilities of the institute in order to carry out the impact sled testing. In particular Wing Commander David Anton for his invaluable guidance, technical assistance and experience in the field of impact biomechanics. Thanks are also due to Mr. Les Neil for his help in building the test fixture and carrying out the impact tests. In addition for his work in developing a test fixture in which to calibrate the sliding knee potentiometers. I am also grateful to Andrew Belyavin for his help and assistance in the statistical analysis of the impact testing results. My thanks also to Surgeon Commander Peter Waugh for his help in smoothing out the computer program in order to manipulate the data from the sled tests.

H W Structures of Leamington Spa, Warwickshire generously carried out the computer validation studies as well as other occupant modelling simulations. In particular I would like to thank Raf Haidar for producing Report no. 5403.

This research was supported in the main part by the Medical Research Council as well as other bodies outlined in Appendix 1. However it is true to say that without the generosity of the Institute of Aviation Medicine this project would not have been possible.

Finally this thesis is dedicated to those occupants of Boeing 737-400, G-OBME on January 8 1989.

ABBREVIATIONS USED IN THE TEXT

%	percentage
AO	Arbeitsgemeinschaft fur Osteosynthesefragen
ATD	Anthropomorphic Test Device
ANOVA	Analysis of variance
ave	average
CAA	Civil Aviation Authority
cm	centimetre
DF	degrees of freedom
FAA	Federal Aviation Authority
Gx	Acceleration in horizontal plane
Gy	Acceleration in a lateral plane
Gz	Acceleration in a vertical plane
kg	kilogram
kN	kilonewton
mm	millimetre
N	Newton
Nm	Newton metres
p	Significance value
std	standard deviation
x^2	Chi-square test

Chapter 1

Introduction and Aims of Study

"An accident is a brief and unforeseen phenomenon. The problems of occupant safety cannot be solved in a satisfactory way without an approach requiring the use of models of the living human being in biomechanical experimentations. The accident victim, although he is not an experimental model, must be studied carefully because valuable information regarding injuries and their causation can be gained."

(From Chapon 1984)

On 8 January 1989 at 8.26 pm a Boeing 737-400 airliner, en route from Heathrow to Belfast, crashed short of the runway at East Midlands Airport and onto the M1 motorway near Kegworth. There were 126 passengers and crew on board, of whom 39 (31%) died at the scene, leaving 87 (69%) injured survivors, who were transported to and treated in hospitals in the Trent Regional Health Authority.

The impact forces were high and resulted in destruction of portions of the airframe but fortunately there was no post crash fire. It was observed by those treating the survivors that a large number had survived albeit with many severe injuries, in particular to the lower limb and head.

It became clear that this accident was on the borderline of survivability and afforded a unique opportunity to investigate the causation of injuries and to study the impact biomechanics.

Regardless of what the cause of an aircraft accident may be, the aircraft occupants will always be affected in some way. Because of the high velocities attained by the aircraft, the forces involved in the crash are violent and in many cases the passengers will suffer serious and frequently fatal injuries.

The commonest injuries in the M1 aircrash were to the pelvis and lower limbs. The range of injuries seen might not have been wholly explained in terms of individual variation, due to age, sex, weight, height etc. Other factors that have been recognised to determine the survivability include the characteristics of the impact pulse eg. duration; peak and rate of onset; restraint system design; orientation of the impact vector relative to the occupant; and seating which can distribute loads over the body and absorb energy. If the occupant's seat and restraint system do not preclude secondary impact of the occupant with the interior of a passenger compartment, then the ability of the cabin interior to distribute the impact load becomes important.

The forces involved following the accident of the Boeing

737-400 (G-OBME) were high and resulted in severe damage to regions of the aircraft. In the areas that sustained severe damage normal seating arrangements were disrupted and mortality was high. In those regions in which the fuselage remained intact, with seating and restraint systems maintaining their integrity, survival was high. However severe injuries were sustained by these occupants not only as a result of the primary forces involved in the accident but also as a result of secondary impacts and interactions with their surroundings.

Limb injuries are a major cause of impairment and disability in victims of impact trauma, they require considerable medical time and resources for their treatment and often have long term sequelae, such as deformity, stiffness and arthritis. In addition lower limb injuries will severely hinder a passenger's ability to escape in the event of a hazardous situation, such as a post-crash fire.

Investigation of the impact biomechanics, in relation to injuries to the pelvis and lower limbs for occupants involved in an aircraft accident, will increase our knowledge and understanding of injury mechanisms and will provide information that will enable more effective impact injury protection systems to be developed in the future.

AIMS OF STUDY

This study was designed to test the following null hypothesis:-

1) That the position adopted at the time of impact did not influence the pelvic and lower limb injuries sustained.

2) That the mechanism of pelvic and lower limb injuries in impact trauma sustained as a result of aircraft accidents was similar to that experienced in automobile accidents. Of particular interest was the instrument panel syndrome or knee-femur-pelvis complex recognised, as a result of automobile research, in causing injuries to the knee, femur and hip.

In addition the study was designed to:-

3) Correlate clinical patterns of pelvic and lower limb injuries sustained by victims of the M1 Kegworth aircrash with a) the structural damage sustained by the aircraft, b) data obtained from anthropomorphic dummy testing and c) the analysis of injury mechanisms using a computer model.

4) Relate the findings from these studies to the development of impact injury protection systems, in

order to decrease the occurrence of pelvic and lower limb injuries following an aircraft accident.

and 5) to use the information obtained to validate as far as possible the computer model used in the study.

Three experimental techniques have been used in this study to investigate the impact biomechanics of occupants involved in the M1 Kegworth aircraft accident:- a) a detailed clinical review of the pelvic and lower limb injuries in the occupants, b) crash testing using a linear decelerator track and anthropomorphic dummies, and c) the use of a computer model.

General layout of the thesis

This thesis will be divided into a number of chapters. The initial chapter reviews the science of impact biomechanics and describes the techniques available for investigating impact accidents.

Essential to the development of this thesis was the M1 aircraft accident, the details of which are laid down in Chapter 3. It highlights not only the crash but also the general injuries sustained by occupants and the use of injury scoring in the investigation of impact aircraft accidents.

In Chapter 4 a detailed clinical review of the pelvic and lower limb injuries in all occupants is reported. The second half of the chapter considers in more detail those injuries sustained by individuals seated in the intact central section of the aircraft.

In Chapter 5 the impact biomechanics of the pelvis and lower limb are investigated using the experimental technique of impact testing using a deceleration sled facility and anthropomorphic test devices.

In Chapter 6 the findings from the clinical reviews and impact testing have been amalgamated and the possible mechanisms of injury to the pelvis and lower limbs, as sustained by occupants on board G-OBME, are explored.

Chapter 7 and 8 describe the use of mathematical computer models in the investigation of occupant kinematics. In Chapter 7 the model has been validated using data generated in this thesis and in Chapter 8 the value of such computer modelling is highlighted.

The final chapter draws conclusions from all the research findings and the implications of this study are discussed together with the proposals for further research.

Chapter 2

Impact Biomechanics

2.1 Introduction

Biomechanics is the study of human response to a variety of loads applied to the body (Fung 1981 and 1985, King 1985). Impact biomechanics is the branch of biomechanics concerned with the response of a body to impact forces and acceleration environments (King 1985). Impact injury to the human body occurs when an external force causes deformation of biological tissues beyond its recoverable limit, resulting in damage to anatomical structures or alteration in normal function (Viano et al 1989).

Transportation accidents are the most important cause of injuries resulting from abnormal loads and accelerations applied to the human body and are a leading cause of death before the age of forty. In many cases the survivors of accidents do not completely recover; sequellae of injuries sustained may result in disabilities and impairments for the patients and cost society a great deal of money (Aldman and Chapon 1984, Viano et al 1989, Trunkey 1983, O'Neill 1985, States 1986, Baker 1984). As the demand for travel grows, road, air and rail networks come under more pressure and the risks of having an accident increase (Bull 1983, Thornley 1990).

The aim of research in the field of impact biodynamics is thus to establish qualitative and quantitative relationships between mechanical forces that develop in an accident and the resulting body damage (Mohr 1978). By applying this knowledge it may be possible to prevent accidental injuries (Aldman 1983, Fung 1981, 1985). Two methods exist:- a) prevent accidents from happening, b) try and influence the accident sequence in such a way as to reduce the risk of injury to people involved in an accident.

2.2 History of Impact Biomechanics

The science of impact biomechanics developed from early observations of natural phenomena. It has long been known that structures that maximise trauma are hard and concentrate loads, such as spears and clubs, whilst conversely shields and armour absorb and distribute loads and protect vulnerable parts of the anatomy (Fung 1981, Mackay 1984).

Modern impact biomechanical research however has developed as a result of observations made by Hugh De Haven during the First World War (De Haven 1969, Snyder 1975, Mackay 1984, Chandler 1990). Following a mid-air collision in which he survived and the pilot died, De Haven attributed his lucky escape to the fact that his cockpit remained structurally intact and he was adequately restrained by a safety harness that protected him from localised contacts and therefore catastrophic injuries. He also noted that his own serious abdominal injuries were related to the buckle of his harness. As a result of his early observations and subsequent interest in injury biomechanics in automobile accidents, De Haven subsequently established the Automotive Crash Injury Research Program.

During the period of the First World War it was observed that more than half of the injuries sustained as a result of aircraft crashes were caused by the aviator striking his head against the sharp cowl of the aircraft (Chandler 1971, 1985(a), Snyder 1975). A simple modification to the cowl practically eliminated head injuries from this cause.

Unfortunately the end of the First World War appeared to end the initial concern over crash related injuries and few major improvements were made in the period up until the Second World War. However with the advent of the Second World War interest was rekindled as a result of pilot shortage. With the development of high speed aircraft the problems of leaving a disabled aircraft became apparent and ejector seats were developed (Chandler 1971, 1985, Mackay 1984). Essential to this research was the question of tolerance of pilots to loads imposed by ejection seats. Arno Gertz (1944), of the Heinkel Aircraft company carried out the first research study into "Biomechanics of Impact" when he investigated the biomechanics of the spinal column following ejection (Ruff 1950, Chandler 1985(a)).

Similar studies were conducted at the Royal Air Force Institute of Aviation Medicine. Techniques were developed in order to simulate the forces experienced by pilots in making ejections. These early impact test facilities included swing seats, acceleration towers, drop towers, acceleration tracks and deceleration tracks (Chandler 1971). These facilities served as a basis for the design of modern impact test facilities. From these early investigations, many on human volunteers, investigators were defining forces that could be tolerated without injury, provided the correct seat design and restraint design were used.

After World War Two research into escape from high speed aircraft continued, although at a much slower pace. Seating and restraint system research evolved as a result of studies directed toward the development of human tolerance data or practical hardware, and was largely aimed at improving occupant protection in automobile accidents (Viano and Stalnaker 1980).

Stapp in the post war years investigated human tolerance to impact and windblast and it became immediately apparent that restraint systems exercised a great deal of influence on the ability of a human to withstand injury (Stapp 1971). It was during this time that he observed that the United

States Air Force lost nearly as many men in fatal automobile accidents as in aircraft crashes, and he began a car crash study using salvaged automobiles. In 1955 interest was expressed in his work by the Society of Automotive Engineers and as a result of this interest annual meetings have been held since: The Stapp Car Crash Conference.

Severy and Mathewson in the 1950's developed techniques of experimental crash testing with instrumented dummies and high speed film analysis (Severy and Mathewson 1954, Mackay 1984). As a consequence of these developments mechanisms of injury causation had been determined by the mid 1960's. As a result methods to reduce the forces and accelerations applied to car occupants in an accident, had been extensively investigated. This eventually lead to the introduction of seat belt legislation and other legislation regarding the design of automobiles (Aldman 1962, Garrett and Braunstein 1962, MacKay 1984, Chandler 1985(a)).

In the aviation industry the majority of research carried out into impact biomechanics following an aircraft accident was directed towards the military aviation community. Groups involved in this work included The Crash Injury Research Project (CIR) in the 1940's. In 1963 their name was changed to Aviation Safety Engineering and Research

(AvSER) (Chandler 1990). AvSER carried out full scale crash tests using large transport type aircraft. The National Advisory Committee for Aeronautics (NACA) also carried out full scale crash testing on small and mid sized aircraft in the 1950's (Chandler 1990).

The progress of both the AvSER and NACA in military crash worthiness resulted in the 'Crash Survival Design Guide' in 1970. However R.F. Chandler (1990) comments that there was 'an apparent lack of progress in civil aviation'.

In 1971 R.F.Chandler working for the Federal Aviation Authority, Civil Aeromedical Institute (CAMI) initiated a program of dynamic testing for seat and restraint system design for the civil aviation community. This culminated in improved seat design and restraint systems in civil aircraft, and the eventual introduction of dynamic loading criteria for passenger seat design.

2.3 Impact Biomechanics

The field of impact biomechanics can be divided into three main areas of interest: injury mechanism, mechanical response and levels of tolerance (King 1985, Viano et al 1989).

2.31 Injury Mechanisms

The mechanism of injury is a description of the mechanical and physiological changes that have resulted in anatomical and functional damage following impact trauma. This is fundamental to injury biomechanics as it provides a basis for determining appropriate measures of response and tolerance to impacts on the various parts of the body (Viano and Stalnaker 1980, Viano et al 1989).

Deformation of tissues beyond their recoverable limit is the general injury mechanism associated with blunt impact (Nahum and Melvin 1985, Viano et al 1989). This mechanism is measured in terms of strain, which is a change in dimension as a result of an outside load. The three types of strain that damage tissues are tensile strain, shear

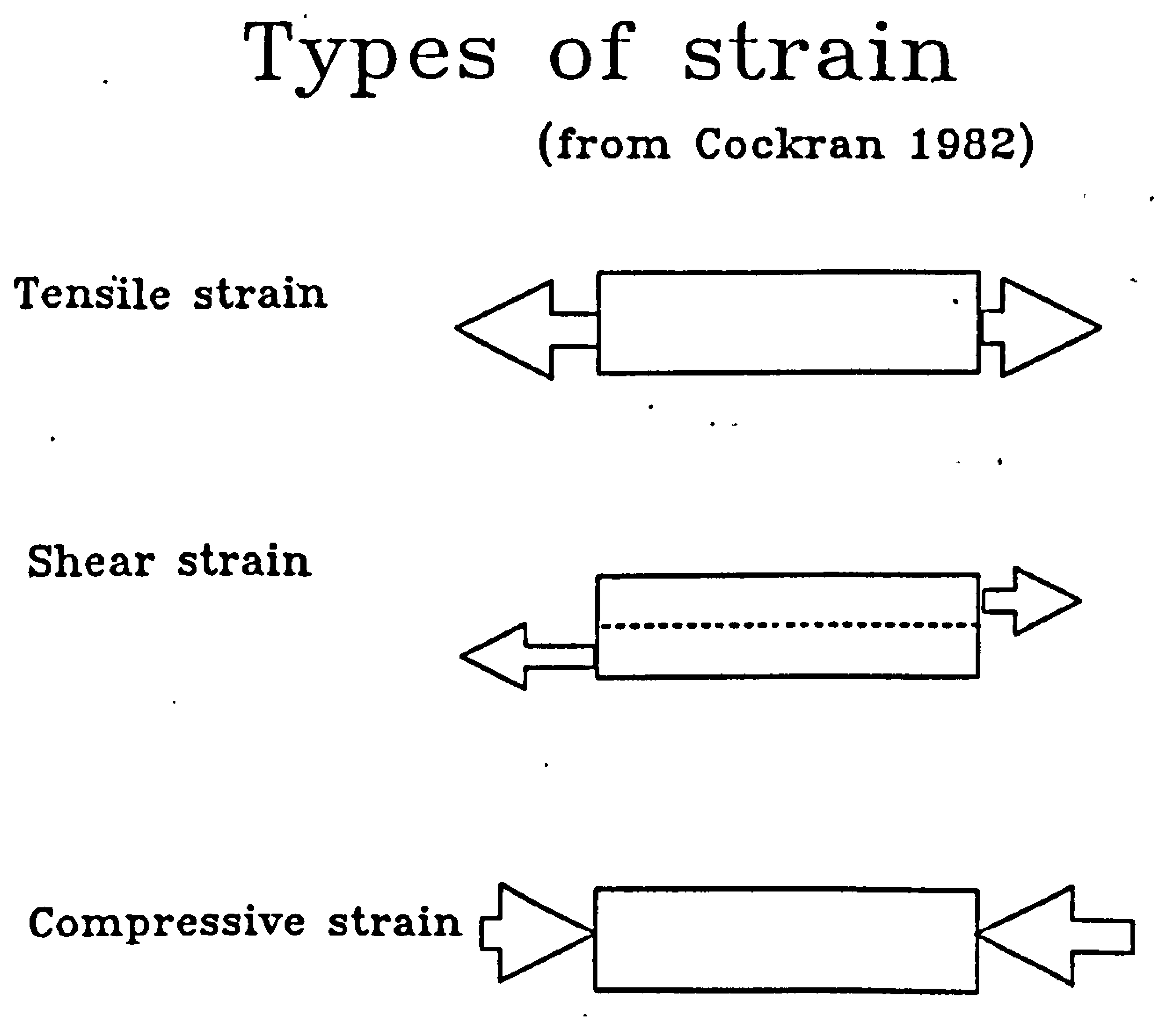


Figure 2.31.1

strain and compressive strain (Perrone and Aniker 1972, Fung 1981, Vogler 1985, Viano 1986, Viano et al 1989). Tensile strain represents an increase in the length of a line drawn on a body: Shear strain represents a change in the angular relationship of two lines drawn on a body: and compressive strain represents a decrease in the length of a line drawn on a body (figure 2.31.1 from Van B Cochran 1982).

An example of how these forces act is given: Impact along the axis of a femur causes an increase in its natural curvature. This results in a tensile strain on its anterior surface and a compressive strain on its posterior surface. A fracture of the femur will occur when its tensile strain limit is exceeded (Cheng et al 1984, Viano 1980, Viano and Stalnaker 1980, Nahum and Melvin 1985, Viano et al 1989).

Shear strain occurs when opposing forces act across a tissue, moving in opposite directions. When the resistive limit is reached the tissue will then fail. This mechanism is important in the causation of head injuries as well as other visceral injuries. This mechanism also explains laceration injuries as well as contusions. In the case of contusion the effect of shear is to damage small vessels beneath the skin (Viano et al 1989).

The rate at which a load is applied is also of importance in the production of injury. If an organ is loaded slowly, much of the energy can be absorbed by deformation without tissue damage. However if loaded rapidly, the organ will fail because it can not deform quickly enough (Lau and Viano 1986). Compact bone also exhibits rate sensitivity during impact. The axial (longitudinal) load a femur can withstand increases with impact velocity, but the strain (bending) at failure decreases (Viano 1980, Viano and Stainaker 1980, Viano et al 1989). This is relevant to design of occupant protection systems.

2.32 Orthopaedic Fracture Patterns

Figure 2.32.1 demonstrates orthopaedic fracture patterns

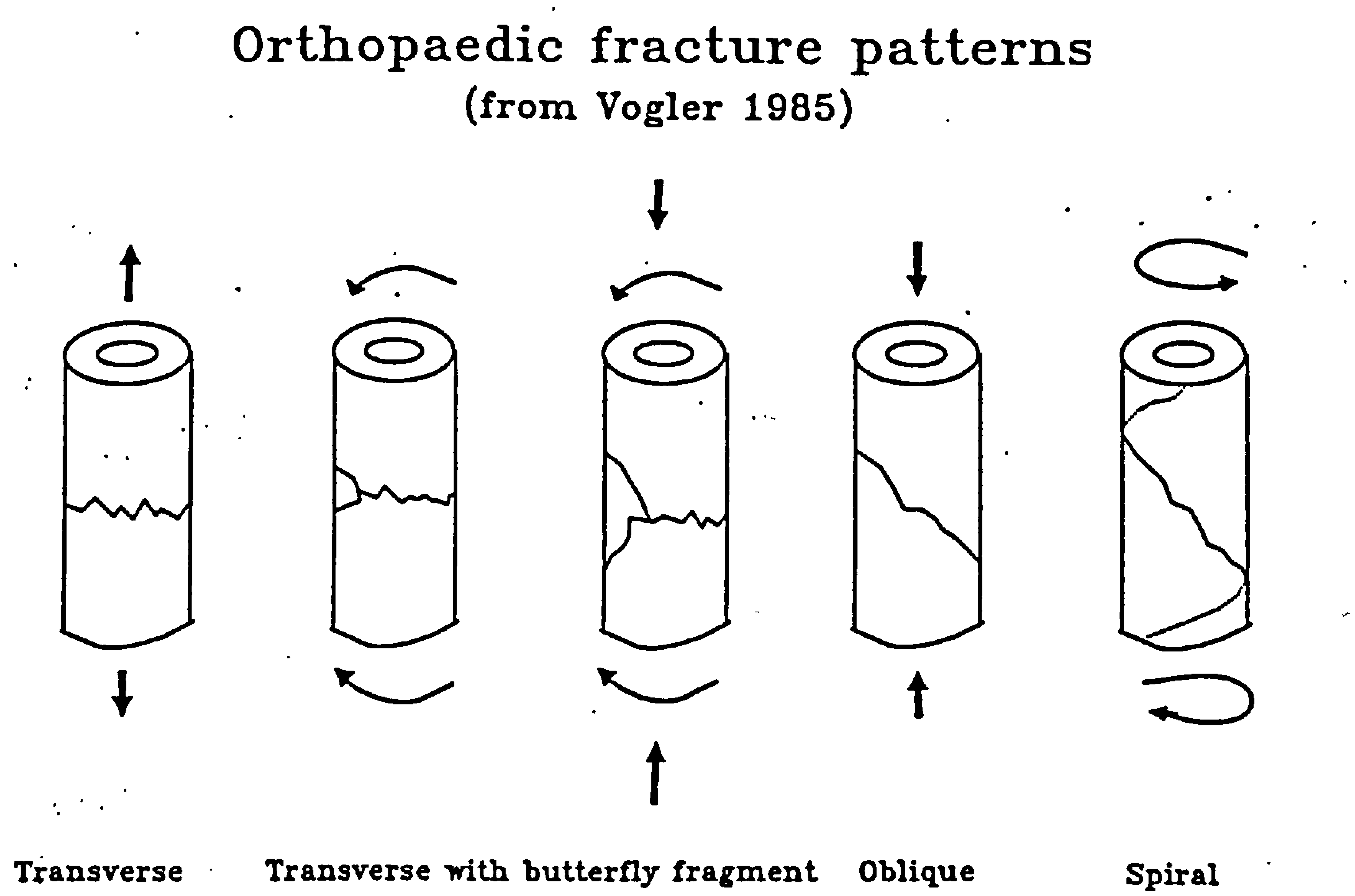


Figure 2.32.1

(from Vogler 1985, and Carter 1985). Fracture patterns can often indicate the mechanism by which the bone was broken.

Transverse fracture

This is a tension loading failure that is initiated on the convex surface of a long bone and is caused by bending or flexural load (Vogler 1985, Carter 1985, Levine 1986). The 'butterfly' fragment transverse fracture is a variation of this with the added element of compression in conjunction with bending. The butterfly occurs on the side of the bone which is in tension.

Oblique fracture

This fracture pattern is produced by a combination of compression and torque. The fracture orientation is a result of induced shear rather than primary torque (Vogler 1985).

Spiral fracture

This is a torque type fracture with the fracture line pursuing a helical course. The orientation of the fracture reflects the tension and compression components on opposing surfaces (Vogler 1985).

Comminuted fracture

This failure pattern suggests the presence of two or more fracture planes and thus at least three or more fragments. It is usually seen as a result of high energy transfer with the load concentrated over a small area (Levine 1986). The underlying strain is massive shear and flexural bending (Vogler 1985).

2.33 Mechanisms: General Considerations

It has been stated that the mechanism of pelvic injury, femoral fracture, dislocation of the knee joint and other lower limb fractures or dislocations are generally well understood. Much of the work describing mechanisms of injuries to the pelvis and lower limbs has been a result of work carried out in the automobile industry (Viano and Stalnaker 1980).

Pelvis

Pelvic injuries are caused as a result of external forces that are applied either directly, to the bony structure, or indirectly as a result of transmitted loading via the femora.

Tile (1988) describes pelvic ring fractures as resulting from external rotation, internal rotation (compression from the lateral side) and vertical shear (figure 1.33.1 from Tile 1988). External rotation is caused by a direct blow on the posterior iliac spines or more commonly by forced external rotation of the legs, and produces an open book type of injury.

Internal rotation (lateral compression) may be caused by a direct blow to the lateral aspect of the iliac crest or an indirect force through the femoral head. This produces

Mechanisms of Pelvic Fractures
(form Tile 1988)

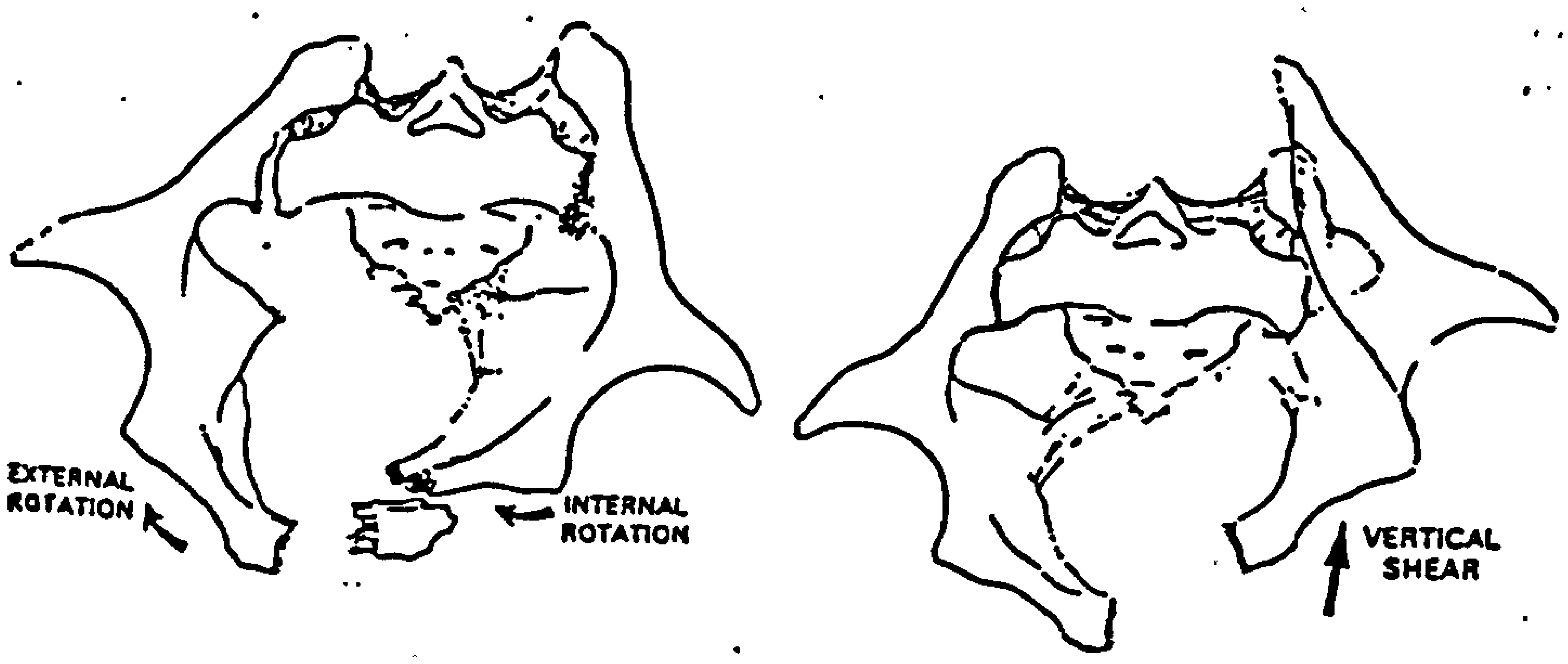


Figure 2.33.1

compression fractures of the posterior complex and fractures of the pubic rami. Pubic rami fractures have been identified as occurring following traumatic lateral impacts in which the load is distributed to the iliac wing and the pelvis as well as the greater trochanter (Dejeammes 1984, King 1985, States 1986, Tile 1988).

Vertical shear forces act across the main trabecular pattern of the pelvis and cause marked displacement of bone and soft tissue. A high incidence of sacroiliac fracture dislocations are seen in aviators as a result of high vertical loads (Mason 1962, Gillies 1965, Hill 1984).

Interpelvic fracture dislocations of the hip occur as a result of a force applied directly to the greater trochanter as in lateral car impacts (Gratton and Hobbs 1969, Epstein 1973, Dejeammes 1984, King 1985, States 1986, Tile 1988, McCoy et al 1989) or as a result of a load applied to the knee with the thigh abducted (Gratton and Hobbs 1969, Dejeammes 1984), seen as part of the instrument panel syndrome or knee-femur-pelvis complex. This mechanism also applies to posterior fracture dislocations of the hip, however the hip is held in flexion rather than abduction (Herman and Epstein 1973). Other interpelvic fractures exist but are uncommon.

Knee-femur-pelvis complex

The response and injuries associated with axial knee impact has long been the subject of numerous experimental investigations, most of which have been aimed at improving occupant protection in automobile accidents (Viano 1980, Viano and Stalnaker 1980, Viano et al 1989).

The following scenario has found wide acceptance:- On impact a seated occupant is propelled forward inertially and the knees strike the dashboard ahead causing injuries to the knee, upper tibia and lower femur (Viano 1980, Viano and Stalnaker 1980, Aldman and Chapon 1983, Cheng et. al. 1984, King 1985, Nyquist and King 1985, Viano and Levine 1986). Posterior cruciate ligament injury is caused by loading of the proximal tibia (below the knee) through the lower dash panel of automobiles (Viano et al 1978, States 1986) displacing the tibia posteriorly beneath the femoral condyles. Impact force is then transmitted up the femur driving it backwards into the pelvis. Femoral fractures result from a bending moment created by axial loading with a possible effect due to the occupants knees sliding under the dash, but with increased energy transfer supracondylar and comminuted shaft fractures are seen (Ritchey et al 1958, Aldman and Chapon 1984, Nahum and Melvin 1985, Nyquist and King 1985, States 1986).

Lower leg, ankle and foot

Injury mechanisms for the lower leg (shin) and ankle are not widely reported in occupants of vehicles involved in impacts (States 1986). Fractures to the tibial plateau, tibia, and ankle have been described as a result of axial loading due to the rearward movement of the toe pan of automobiles, coupled with torsion and / or bending moments (Nyquist and King 1985, States 1986) at impact.

Anderson in 1919 described a common foot fracture in early aviators the 'Aviators astragalus' (Coltart 1952, Hawkins 1970). The mechanism of injury was explained as a result of forces transmitted through the sole of the foot resting on the rudder bar, thus causing hyper-dorsiflexion. This injury is now more commonly associated with automobile accidents (Hawkins 1970, Penny and Davies 1980).

Swearingen et al in 1961 published a paper that has been much quoted by investigators working in the field of crash investigation and biomechanics. It has also formed the basis of many ideas on the causation of injuries following an aircraft accident and for this reason has implications for the design of aircraft interiors. The purpose of the paper was to 1) present a detailed description of the areas which are traversed by the human head, trunk, and appendages during flailing motions in a crash, by an occupant restrained with a lap safety belt restraint only, 2) relate present aircraft cockpit and seating arrangements with these areas of motion, 3) present an analysis of aircraft injuries, and finally 4) discuss some of the body impact forces which may be involved in a typical survivable transport crash.

Figure 2.33.2 demonstrates the flail area in which the body is free to move during actual crash impacts, and is described within the upper two-thirds of a sphere 10 feet in diameter. As can been seen the body "jack knifes" or flexes over the lap belt and the limbs flail forward. The occupant restrained by a lap belt experiences a deceleration in a forward and downward direction (Fryer 1965). The head and torso are the usual sites of fatal lesions following aircraft accidents (Swearingen et al 1961, Mason 1962, 1973, Gillies 1965, Stevens 1970, Hill

The Flail Envelope

(from Swearingen et al 1962)

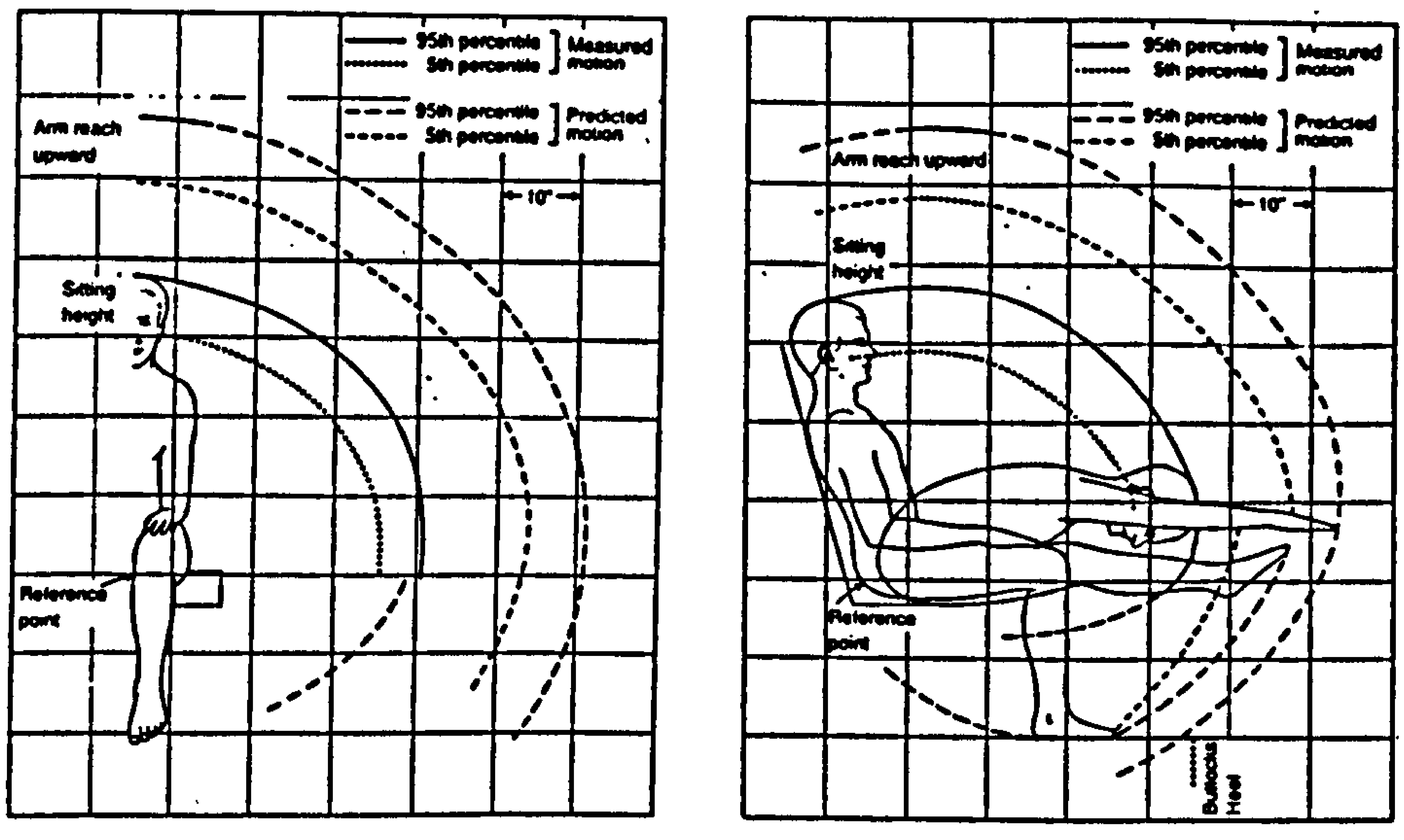


Figure 2.33.2

The Lethal Area

(from Swearingen et al 1962)

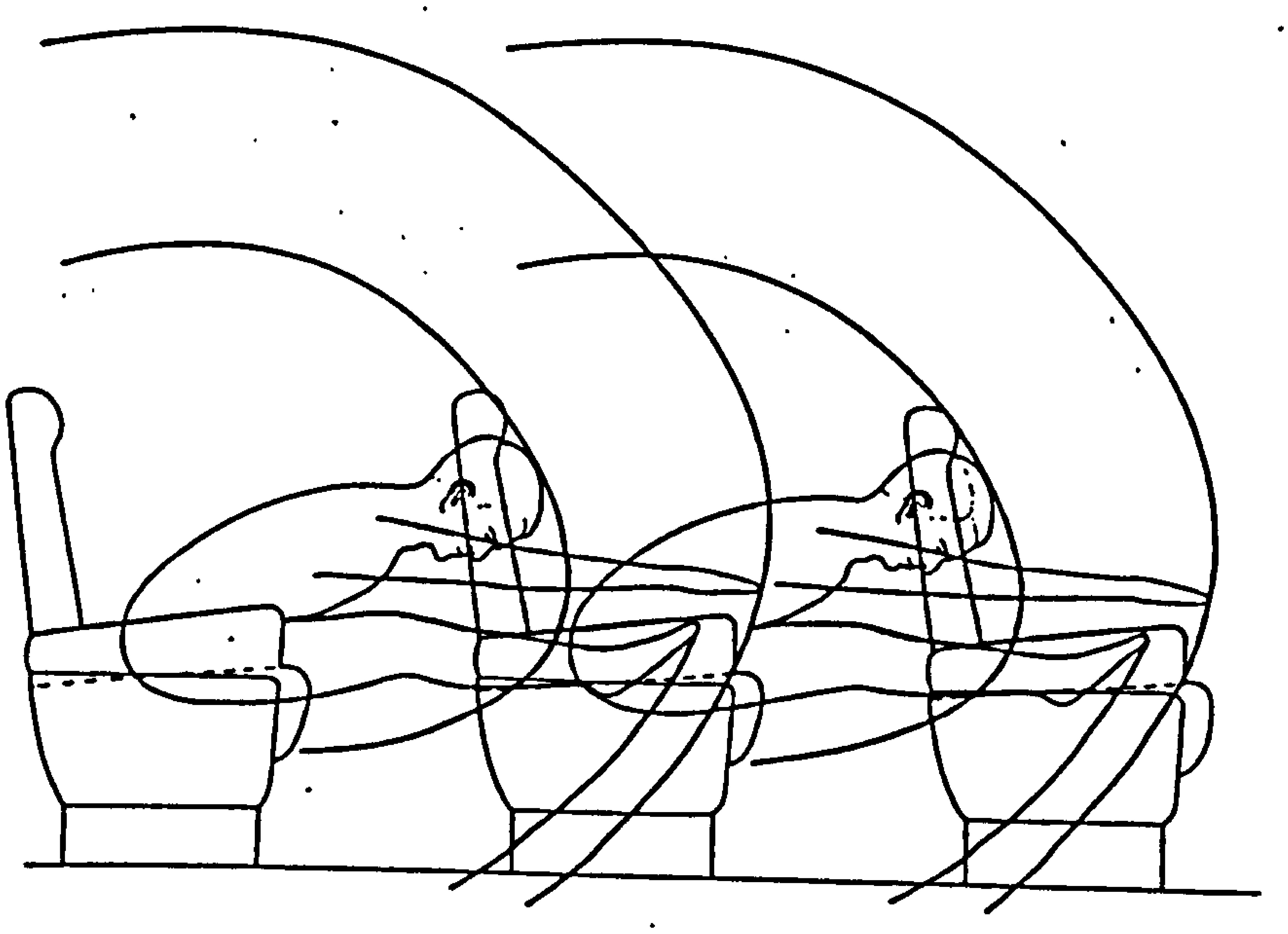


Figure 2.33.3

1982, 1984) and the area within the head clearance curve may be considered the lethal area (figure 2.33.3). In addition the area swept over by the arms and legs might be classified as the "incapacitating" area. Unconscious and incapacitated passengers in commercial transport accidents are often trapped and die in the smoke, fumes and fire that follow a crash (Swearingen et al 1961).

Injury mechanisms for lower limb and pelvic injuries will be further discussed in relation to the clinical review of pelvic and lower limb injuries sustained by passengers and crew involved in the M1 Kegworth (G-OBME) aircraft accident on 8 January 1989.

2.4 Biomechanical Response

Once an injury mechanism has been described the next step is to quantify the biomechanical response during the impact. The measurement of biomechanical response should characterise how an organ or tissue reacts to deformation, or how the inertial resistance of the body or tissue responds to an applied load or motion. This information can then be used to analyse the injury process and to develop mechanical human surrogates or mathematical models that behave in a human like manner under impact conditions.

Mechanical human surrogates or mathematical models will not be considered further in this section as they are not biological models and cannot therefore be used to establish relationships between mechanical parameters and injuries. However other methods exist to determine biomechanical response and these will be discussed.

2.41 The Human Volunteer

Human volunteers are clearly the best experimental models available to determine the biomechanical response to an applied deforming force (Kazarian and Von Gierke 1978, Hill 1984, Chapon 1984, Viano et al 1989). Strict guide lines exist limiting the acceleration levels that can be used with volunteers. However much early work in impact biomechanics have used extensively the human volunteer or indeed the investigator. Only one man-rated impact facility exists in England (Anton 1990), at the RAF. Institute of Aviation Medicine, Farnborough., this being a military establishment. This unfortunately results in data that reflect the response of fit young males, information that may be most applicable to the military environment.

Further limitations of this approach include the problem of subjecting the volunteer to levels of impact that will not cause injury. As this may vary from individual to individual it is ensured against by starting off at low levels of impact acceleration, gradually increasing until the subject feels pain or discomfort (Hill 1984, Chapon

1984, King 1985). Because volunteers are young and fit and impact levels are lower than the forces occurring in a crash situation, muscular activity of the volunteers can modify their dynamics (Begman et al 1980, Chapon 1984). However in an unexpected impact situation a relaxed volunteer's reflex responses are too slow to have a significant effect on loads and accelerations sustained (Begman et al 1980).

In spite of these limitations volunteer experiments can provide useful information on kinematics at low impact levels. This information unfortunately cannot be extrapolated to high impact levels.

2.42 The Accident Victim

Ideally the human response to an impact would be obtained from living subjects under various crash conditions. Unfortunately because of the unforeseen nature of accidents, individuals involved in an accident cannot be instrumented with electronic measuring devices (Viano et al 1989) which will measure and define the biomechanics of injuries sustained. The accident victim however provides an important catalogue of injuries sustainable as a result of impact trauma.

Clinical studies and autopsy reports of victims of accidents are potentially the most valuable source of material to investigate injuries and their cause. Much early work on impact biomechanics has evolved from these observations (Hasbrook 1957, Mason 1962, 1968, 1973, Fryer 1965, White 1966, Stevens 1970, Kirkham 1982). However Snyder (1975) and Hill (1984) argue that even though a lot of valuable information has been gained, if the number of accidents that occur annually is considered only a few have been properly analysed, and rarely is an attempt made to correlate the injuries received with the causation of the injury. Information relating specifically to injuries to the extremities is seldom reported, even though this information may be crucial to the determination of why the occupant failed to escape following a crash (Snyder 1975).

If the forces and parameters involved in accidents can be accurately determined then the accident victim may become increasingly important in the investigation of injury biomechanics.

2.43 The Human Cadaver

Although measures of response to non injurious impact can be obtained from volunteer experiments the primary data on impact response at injury levels must be obtained using human surrogates: Human cadavers and or anaesthetised animals.

The human cadaver has both limitations and advantages. Morphologically the cadaver is an identical model to the 'living' subject, and is a suitable research model to simulate gross geometric and material properties of the human or to study the mechanical response of a body segment (Kazarian and Von Gierke 1978, Chapon 1984, Hill 1984, Viano et al 1989). However since the cadaver is no longer a functioning biological system, injury to soft tissue, hollow and parenchymatous organs can only be inferred. Muscle tone no longer exists and as a result energy transmission and attenuation, in both hard and soft tissues, cannot be precisely defined (Kazarian and Von Gierke 1978). Unfortunately most cadavers tend to be of advanced age with pre-existing illness and osteoporosis, which further effects the biomechanical responses of the tissues.

Embalming cadavers may effect the biodynamic properties of the tissues although these affects are much debated. The lack of an intact circulation means that many signs of trauma will not be evident. Such signs as bruising or abrasions may indicate a moderate injury of importance in the setting of safety design criteria (Hill 1984). Further the lack of muscle tone, with limpness or rigor mortis of the cadaver, prevents easy manipulation and significantly effects the kinematic behaviour (Chapon 1984, Hill 1984).

Despite these problems the human cadaver is a useful model and has been extensively used in the evaluation of the tolerance of bone to deforming forces. The design of mechanical human surrogates (Anthropomorhic test devise (ATD) or 'dummy') that behave in a human like way owes much to data derived from cadaver studies (Foster et al 1977, Mertz 1985).

2.44 The Animal

The cadaver as a surrogate tends to be of advanced age, has no active muscular response that may influence its realism at lower levels of acceleration, and cannot be used to assess functional changes due to injury. The only way to obtain information on the physiological responses to injury is to subject live, anaethetized animals to experimental impact. Even their physiological responses are influenced by the anaesthetic (Kaserian and Von Gierke 1978, Chapon 1984, Viano et al 1989).

Being a living model, it can be subjected to high impact levels likely to produce severe injury. The animal model is not a human and variations in anatomy and physiology may alter the response to impact trauma in a way that makes extrapolation of findings to the human non valid. However animal studies are critical to the study of the brain and spinal cord injuries, arrythmias and shock (Viano 1989), but their use is limited in determining the quantitative values of human tolerance to impact.

2.5 Impact Tolerance

At some measurable level, tissues involved in impact trauma will not be able to recover. Human tolerance to impact injury is defined in terms of the threshold that is selected (King 1985).

Safety engineers, in order to effectively design impact protection devices, need to know what forces and loads the body can withstand. Injury criteria are set by legislative bodies in order to offer guide lines for engineers designing seats and restraint systems. Federal Aviation Administration Advisory circular no. 21-22 defines suggested numerical values for aircraft use (Pontecorvo 1985). Injury criteria must obviously take into account the impact circumstances. Criteria for lower limb injury laid down for car manufacturers may indicate a level of injury consistent with minor injuries. This may be acceptable in a car accident, but in a burning aircraft following an accident lower limb or head injury could have serious implications for the occupant.

Two techniques have been used to investigate the tolerance of the human body or isolated tissues to impact trauma:- a) dynamic methods and b) static methods. Dynamic testing attempts to simulate real crash conditions, such as would be experienced by the victim of an accident, and uses impact test facilities (Chandler 1985(b), 1987). Static testing uses isolated tissue, or parts of bodies to which loads are applied until they fail. London (1977) has criticised static testing as being artificial and bearing no resemblance to reality.

A number of different ways of defining tolerance have been used. The types and grades of tolerance now accepted are as follows (Hill 1984. King 1985, Viano et al 1989):

a) **Voluntary tolerance**

This is the lowest level of tolerance and is sometimes referred to as the 'ouch' level. It is defined as that level of impact force which someone can withstand voluntarily, without sustaining injury. It is extremely variable and injury is not the end point of these tests, although injuries are occasionally sustained by the volunteer but these are usually of a minor degree.

b) **The injury threshold**

This represents a level just below which injury occurs to a given organ or tissue and is usually reached accidentally.

c) **Moderate injury**

Complete recovery from any injury produced without any residual impairment of function. It is at this level that injury criteria are laid down for the design of injury protection systems.

d) **Severe injury**

This may be defined as the degree of force needed to produce injuries that are not fatal or the level at which fatal injuries begin to occur.

e) **Fatal injuries**

This is an impact level in which fatal injuries occur. These higher levels of tolerance are determined from tests on human surrogates, such as human cadavers or animals. In the evaluation of the performance of protection systems, mechanical surrogates are used (anthropomorphic test device) at these high impact levels. Instrumentation of these devices allow measurements to be made that can be correlated with injury criteria and therefore injury severity of a body segment (King 1985, Pontecorvo 1985).

2.51 Factors Influencing Tolerance to Impact Trauma

Early studies in impact biomechanics attempted to determine whole body tolerance of occupants who were restrained in a seat. Other early investigators attempted to estimate the G level encountered by individuals who survived falls from great heights. (Dehaven 1942, Snyder 1963, 1971). More recent work has established the tolerance of individual regions which when injured may pose a threat to life or result in long term disability. Regional tolerances are therefore more useful in assessing the injury potential of protective systems or restraint devices.

Investigation of falls from a great height has highlighted factors important in the causation of impact trauma as well as demonstrating the high forces that are survivable. Factors influencing response to impact trauma are reviewed by Snyder (1963, 1971) and can be split into two groups : Physical factors and Biological factors.

<u>Physical factors</u>

i) **Orientation of the body**

Since tolerance is related to the direction of impact, a unified reference system has been adopted and proposed by the Biodynamics Committee of the Aerospace Medical Panel, AGARD, (1961). Figures 2.51.1 and 2.51.2 represent the convention of signs used. Head first (+Gz) falls are the

Linear Acceleration

Acceleration Description	Physiological Standard	Vernacular Description
Forward Accel.	+Gx	Eyeballs in
Backward Accel.	−Gx	Eyeballs out
Headward Accel.	+Gz	Eyeballs down
Footward Accel	−Gz	Eyeballs up
R. Lateral Accel.	+Gy	Eyeballs left
L. Lateral Accel	−Gy	Eyeballs right

Figure 2.51.1

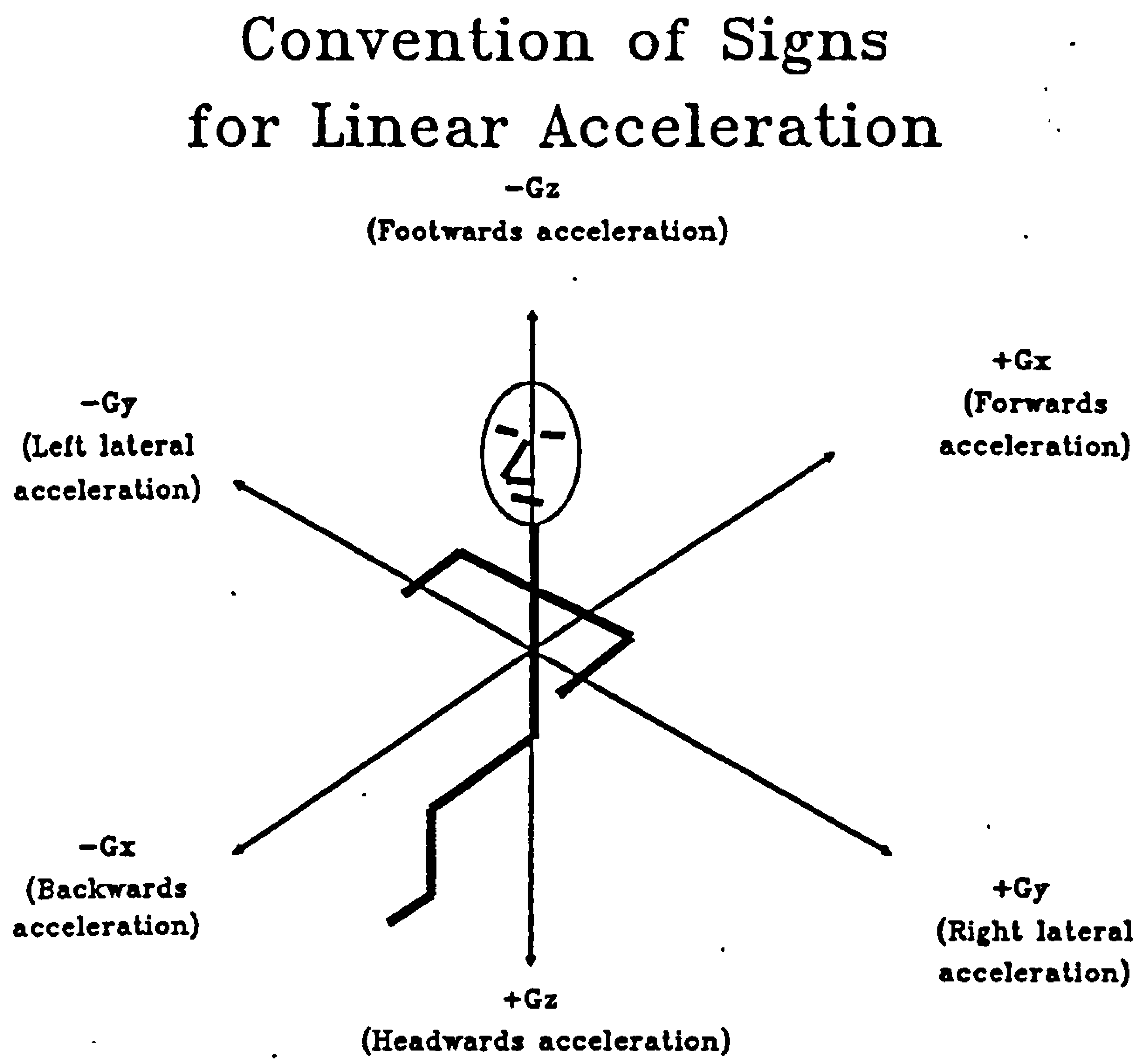


Figure 2.51.2

least well tolerated. In feet first (+Gz) impacts severe injuries are seen in the feet, ankles and lower limbs. In seated impacts (+Gz), pelvic and vertebral injuries prevail. In side impacts (±Gy) the upper extremities, thorax and vertebral column, followed by pelvic injuries are most commonly involved.

ii) **Magnitude of acceleration**

The magnitude of acceleration is generally expressed in G units and 'rate of onset'. G represents the acceleration due to gravity which we sense as weight, thus a pilot exposed to an acceleration of six times that of gravity

(6G), will have experienced his body weight increasing six fold (Harding and Mills 1988).

Simplistically it is true to say that low G levels may be easily tolerated and at high G levels serious injury and death will result. This however is complicated by the effect of 'the rate of onset'.

iii) Distribution of force

Dehaven (1942), in his study of free falls concluded that the greater the area over which a load is applied the smaller the load per unit area, and the greater the survival in free-fall and tolerance of high impact forces. This concept is of great importance in the design of passenger protection systems.

This effect is however modified by attenuation of energy in body tissues. For example it is known that in feet first impacts, as experienced in parachutists, muscle tonus and degree of bending of the legs affect injury tolerance (Snyder 1971).

The design and tightness of restraint systems and the characteristics of seating design will modify the distribution of forces and the manner in which they act.

iv) Material impacted

The relative deformation of the object impacted is of major importance as an injury or survival determinant The degree of elasticity or solidity of a material affects the deformation distance, energy attenuation and time duration of impact. Thus soft muddy ground is obviously preferable to jagged rocks upon impact (Snyder 1963). Likewise in automobiles a padded dashboard is preferable to a sharp metal dashboard (Nader 1965).

v) Time duration of impact

The duration of time that the force has been applied is recognised as one of the most critical factors in human impact tolerance. In impact this generally refers to the time required to reach the peak force at initial impact, but may also refer to the total time that the force is applied.

The longer an impact acceleration is applied, then the greater will be its effects (Dehaven 1942, Snyder 1963, 1971, Hill 1984). For example 45G can be tolerated in a chest to back (-Gx) direction if applied for a 0.044 second pulse. If applied over 2 seconds then considerable injury will result (Hill 1984). Stapp (1961) suggested that if an impact duration is less than 0.2 seconds the tissues behave like rigid materials and damage and failure are independent of gradients of fluid displacement, ie. the tissues do not have time to react.

Biological factors

Individual variations in victims of impact trauma also account for variations in response to impact trauma as well as recovery from injuries. As mentioned previously much work on human tolerance levels has been conducted on young physically fit males. Thus the accident victim or victims involved in falls from a height may provide data relating to individual variation such as sex, age, physical and mental condition, race, pre-existing disease and other biological factors.

2.52 Secondary Impacts

The effect of secondary impact requires special mention. Although the primary impact may be tolerable, death or severe injury may result from secondary impacts. These refer to impacts that are usually a result of flailing (Swearingen et al 1962) of the limbs, torso and head, or failure of the restraint systems. Seating design and restraint systems aim to modify an occupants behaviour in order to prevent serious secondary impact from occurring as well as augmenting the primary forces. If the force of an abrupt deceleration following an impact exceeds the strength of the retaining devices the passenger will be hurled in the corresponding direction sustaining secondary impacts and injuries (Hasbrook 1957, Kreft 1971).

It is apparent that following a sudden deceleration an unrestrained occupant will become a free projectile obeying Isaac Newton's (1642-1727) three laws of motion.

2.53 Injury Tolerances

Weber in 1856 (cited Melvin and Evans 1985, Nyquist 1986) determined static loads required to fracture entire bones by a three point loading mechanism, transverse to the long axis of the bone. The mean loads at fracture for the femur and tibia is recorded below in table 2.53.1.

Table 2.53.1

	Femur	Tibia
Male, kN	5.09	3.06
Female, kN	3.98	2.33
Support distance, cm	18.3	21.6
Maximum moment, male N-m	233	165
Maximum moment, female N-m	182	125

King (1985) investigated the effect of dynamic lateral loading over the greater trochanter of the femur, using an impactor, on cadaver subjects. The most frequent fractures were of the pubic rami followed by fractures of the proximal femur, dislocation of the sacroiliac joint, fracture of the iliac wing and fractures of the acetabulum. These injuries were seen to occur at loads ranging from 4.4 to 12.9kN depending on sex (average 8.6kN males and 5.6kN females) (King 1985).

Tarriere et al (quoted King 1985) in 1979 reported on the outcome of drop tests to the lateral aspect of the pelvis in cadaver studies. Fractures seen were commonly to the pubic rami. They proposed a tolerance level of 80 to 90G for pelvic acceleration. Vertical loading tests to the pelvis, achieved by dropping weights onto the lumbar spine of cadaveric specimens, demonstrated bilateral dislocation of the sacroiliac joints at loads of 3.7 kN (Fasola et al 1955).

Paterick et al (1966) demonstrated pelvic fractures in cadavers, following dynamic axial loading to the femur of 6.23 to 17.1 kN. They proposed a fracture threshold level should be set at 6.2 kN. Melvin and Nusholtz (1980) performed sled tests on unembalmed cadavers and demonstrated a range of injuries to the pelvis and lower limb, including femoral fractures and injuries to the the patella and femoral condyles. Impact loads varied from 8.9 to 25.6 kN, suggesting the proposed fracture threshold could be raised.

Viano and Stalnaker (1980) investigated mechanisms of femoral fracture using denuded cadaver femurs. The femurs were subjected to axial impacts at the knee by an impactor. They described three features of axial knee impacts: knee compression, femoral bending and compression and femoral displacement with load transfer to the hip joint. Each facet resulted in injuries to the individual components. On average condylar and femoral fractures were seen with knee loads of 10.6 ± 2.7 kN. However Nyquist and King (1985) in their review of the 'Lower extremities' comment that injury tolerance data for the upper leg consist primarily of axial loading to the femur, and that impact testing produces a preponderance of distal femoral fractures, a situation not borne out by clinical observations.

Viano et al (1978) produced posterior cruciate ligament avulsions at 5 kN with an impactor that struck the tibial tuberosity. If the impactor spanned the knee joint then the force required increased to 7 kN. However Noyes and Grood (1976) demonstrated that the force required to rupture an isolated anterior cruciate specimen ranged from 622 newtons to 1170 newtons depending on the age of the individual. Patellar fractures can be demonstrated at 2.5kN to 9.0kN, depending on the impactor. In general it can be demonstrated that patellar tolerance is increased by padding and load distribution (Nyquist and King 1985, Melvin and Evans 1985).

Extensive human cadaver testing has been conducted in connection with pedestrian impact injuries. As a result of such research using transverse impacts to simulate bumper injuries, Kramer et al (1973) found they could induce fractures in 50% of tibia tested with a loads of 4.3 kN (Nyquist and King 1985, Melvin and Evans 1985, Nyquist 1986). Axial loading of 5.5 kN to the planted foot produced calcaneal fractures (Nyquist and King 1985, Melvin and Evans 1985, Nyquist 1986).

Injury criteria or loads above which bony injury can be expected, have been proposed as a result of cadaveric studies. Such criteria are important to engineers designing impact protection systems. The Federal Aviation Authority (FAA) have outlined proposed injury criteria in FAA Advisory circular AC no.21-22 (Pontecorvo 1985) and are outlined below. The criteria selected are the same as those used in the automobile industry.

Table 2.53.2

<u>Leg injury</u>		
i) In line with femur	10kN	(2250lbs)
ii) Patella (concentrated load)	2.5kN	(560lbs)
iii) Transverse (lower leg)	4.45kN	(1000lbs)
<u>Spinal injury</u>		
Lumbar compression load	6.7kN	(1500lbs)

To this list can be added a recommended lateral loading tolerance level for the pelvis of 10kN (King 1985).

2.6 Injury Assessment Technology

The Stapp Car Crash Conferences have been held annually since 1955, and have highlighted the improvements made to car design and safety in the last 30 years. It has been demonstrated that blunt impact and acceleration injury can be significantly reduced in the automobile environment through the use of crushable vehicle structures, which absorb impact energy, and restraint systems which allow the occupant to decelerate more slowly with the crushing vehicle. The effectiveness of these protective systems cannot be demonstrated or investigated without some means of simulating a crash situation in a laboratory.

2.61 Anthropomorphic Test Devices

As mentioned previously human cadavers or other biological surrogates have been used in crash simulation, however they are difficult to use, do not provide repeatable information and are unable to be used once an injury has occurred. Mechanical surrogates (anthropomorphic test devices (ATD) or dummy) are therefore used and are able to simulate the human body in respect to mass, shape, size, stiffness and kinematics following an impact (Foster et al 1977, Mertz 1985).

However ATD's are not actually injured, but rather levels of biomechanical response are established that are judged likely to result in injury if the dummy were human. Impact

Hybrid III Anthropomorphic Test Device

(Anterior view)

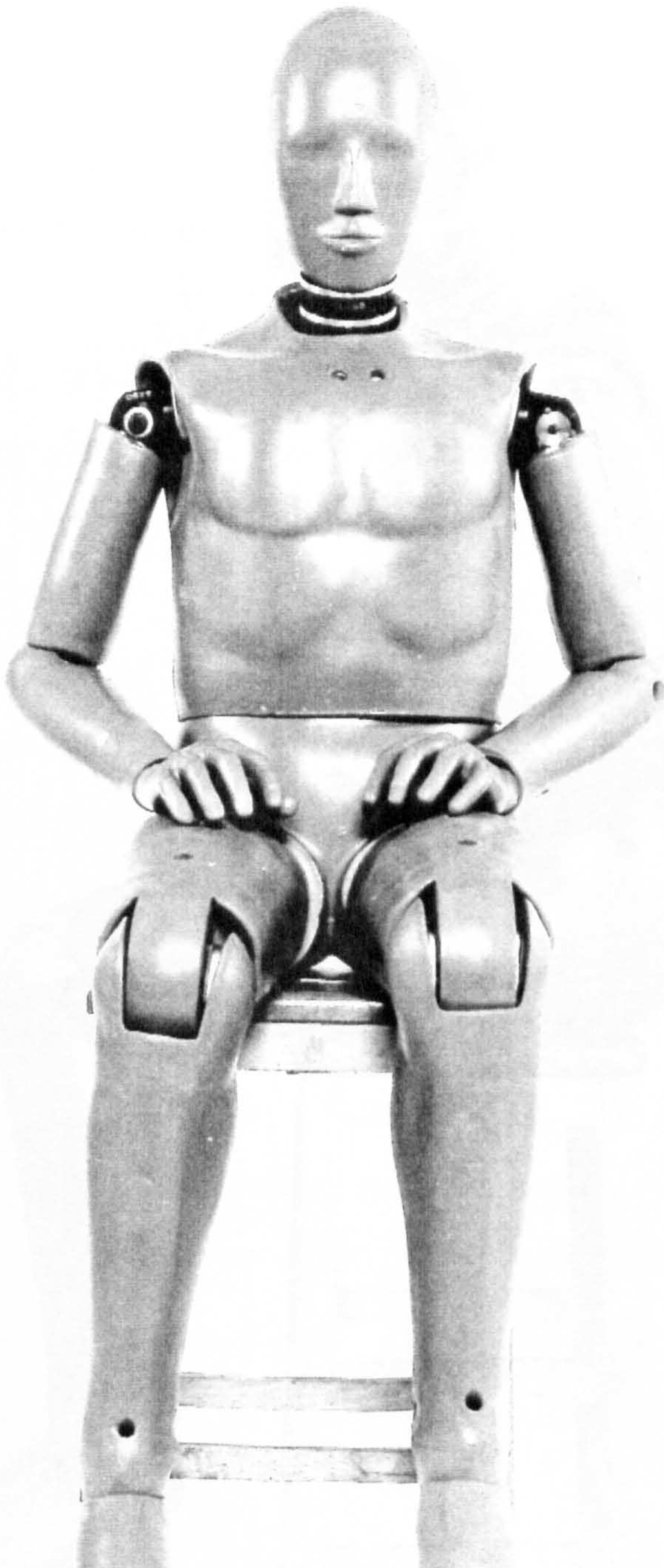


Figure 2.61.1

Hybrid III Anthropomorphic Test Device

(Lateral view)

Figure 2.61.2

injury criteria determined using biological surrogates are expressed in parameters that are measurable on an ATD. Thus using instrumented ATD'S and a range of impact exposures, the risk of injury and disability to the human can be assessed (using injury criteria) in a repeatable manner and improvements made to an occupant protection system.

Anthropomorphic dummies are usually classified according to their physical size. For example, the height and weight of a 50th percentile adult male ATD approximates the median height and weight of the adult male population of the United States (Mertz 1985). Other adult dummy sizes include the 5th percentile adult female and the 95th percentile adult male dummy. Child ATD's are also available.

A number of designs of ATD exist. They can be classified as either "frontal impact dummies" or "side impact dummies". The Hybrid III (figure 2.61.1 and 2.61.2) is a 50th percentile adult male dummy developed by General Motors in 1976 (Foster et al 1977) and is an example of a "frontal impact dummy" that has found wide application in both the automobile industry as well as the aviation industry.

The biofidelity, or the degree to which pertinent human physical characteristics are simulated, of ATD's have been criticised as they cannot simulate physiology and pathology, however they do provide acceleration, deformation, and kinematic data which in most cases can be correlated with human impact response (Foster et al 1977, Mertz 1985, Viano et al 1989).

2.62 Dynamic Impact Test Facilities

An impact test facility is a dynamic facility that is able to produce a controlled impact representative of an actual crash (Chandler 1971, 1985(b), 1987, Gilles 1971, Dutton 1974, FAA Advisory circular 1988). They were developed during World War II as a means of investigating restraint systems, cockpit design and ejector seats in aircraft. A variety of test facilities have been developed and include swing seats, acceleration towers, drop towers, acceleration tracks and deceleration tracks. These test facilities served as a basis for similar devices in use today (Chandler 1971).

None of these facilities however will accurately represent the complex impact conditions which take place in real world aircraft crashes. One major difference is that the dynamic test facility must always start and stop at zero velocity. There must be an acceleration phase and a deceleration phase. In an aircraft crash the acceleration phase is always gradual and usually well separated in time from the deceleration phase. With a test facility the acceleration phase and the deceleration phase are always closely related.

Although impact facilities may be able to simulate a crash pulse, i.e. the magnitude of acceleration change, it is difficult to simulate a change in acceleration direction (acceleration vector) that may be experienced in a crash (Chandler 1971, 1987). Facilities that are able to change the acceleration vector during the time sequence of a crash are unfortunately not available.

2.63 Deceleration Sled Facilities

In an airplane crash the impact takes place as a deceleration, so loads are applied more naturally in test facilities that create the test impact pulse as a deceleration (Chandler 1987, FAA Advisory Circular 1988). Since it is easier to design a test facility to extract energy in a controlled manner than to impart energy in a controlled manner, several deceleration facilities can be found (Gilles 1971, Aston 1971, Dutton 1974, Chandler 1985(b), FAA Advisory Circular 1988).

Problems can exist in such facilities. During the acceleration phase of the test, in which sufficient velocity for the test is acquired, the acceleration may cause the ATD to move from its intended pre-test position. This can be avoided by using a lower acceleration for a relatively long period and providing a coast phase prior to the impact.

Other horizontal test facilities in common usage include acceleration sled facilities; in which the impact pulse is supplied at the beginning of the test, and impact with rebound test facility; in which the impact takes place in the middle of the test with the impact energy returned to rebound the energy and test rig in the opposite direction.

2.64 Mathematical Models

Another test tool used to assess injury risk is the mathematical model. Mathematical models and computer simulations have been developed over the last 20 years in order to predict a body's response to injury producing conditions that cannot be simulated experimentally eg. the multi-directional acceleration crash pulse of real world crashes, and to predict responses that cannot be measured in surrogate and animal experiments (Von Gierke 1971, King and Chou 1976, Ward and Nagendra 1985, Viano et al 1989). Models offer great flexibility. The investigator can vary any parameter in the smallest increment and measure the difference that particular change has over final outcome.

Two types of biodynamic model exist: regional models eg. head or spine; and whole body models. Regional models will not be considered. Whole body models have found wide use as predicative tools of body kinematics and acceleration during impact in the automobile industry (King and Chou 1976) but their use has been extended to other injury conditions. Graphics programs have been interfaced with these models so that body motion can be visualised.

Computer models have however been criticised. Pitfalls include lack of validation, over-sophistication, and a lack of properties of biological tissues to go into the models (Panjabi 1979, Ward and Nagendra 1985).

Validation of computer models relies on correlation with experimental tests. If a models' predicted response comes close to the measured results, the model is assumed to be validated (Kasarian and Von Gierke 1978, Ward and Nagendra 1985, Laananen 1985)

Models of the whole body

Gross-motion simulators are the class of mathematical model formulated to describe the kinematic and dynamic response of a vehicle occupant involved in a collision. Two dimensional or three dimensional models exist (King and Chou 1976, Wismans et al. 1982, Ward and Nagendra 1985). The models are basically computer programs that have been developed to solve displacement and rotation equations developed in simulations. Dimensions and mass of the body points can usually be changed to represent individual accident victims.

Some of the crash victim simulation models have been used for aircraft safety related problems. These include the Articulated Total Body (ATB) computer model, and the Seat/Occupant Model- Light Aircraft (SOM-LA) (Wismans et al 1982, Laananen 1985). The ATB model was developed to simulate aerodynamic forces experienced during ejection, whereas the SOM-LA model was developed as an engineering tool for use in crash worthiness design and evaluation of seats and restraint system for light aircraft.

MADYMO is a crash victim simulator that has been used extensively in the automobile industry in the development of crash safety devices (Wismans et al 1982, Ward and Nagendra 1985, HW Structures Ltd.). It has also been used in biomechanical crash research.

Chapter 3

The Circumstances of the M1 Kegworth Air Accident and Subsequent Events

This chapter reviews the details of the crash of the Boeing 737-400 (G-OBME) aircraft on the 8 January 1989 at 8:26 pm., the actual events which took place, the injuries to the occupants and the consequences of the aircraft accident. Much of this work was undertaken by a team of researchers - the NLDB Study Group (See appendix 1). The author's role was both to carry out much of the investigative work, but also to co-ordinate the research work of all the members of the group.

3.1 The aircraft

Boeing is the world's largest producer of commercial passenger aircraft. The 737 series of passenger aircraft have been in production since 1967 with several updates to cope with the growing demand for aircraft suitable for medium haul travel. As a result of the success of the 737-300 series Boeing introduced the 737-400 series in 1988, this series incorporated improvements to the structure of the aircraft, the control systems, the cabin interior features and the flight deck design to provide an aircraft with increased capacity, improved economics and performance (Air International 1989).

The 737-400 was a "stretched" variant of the 737-300, being some 3 metres longer with modifications to the fuselage, wings, under carriage and engines. It was a variable seat aircraft with a maximum seating capacity of

156 passengers and crew. External features that distinguish it from the 737-300 were the presence of four over-wing escape hatches, two on each side (instead of one on each side).

British Midland Airways purchased two 737-400 aircraft with the 156 seating configuration (G-OBME and G-OBMF) some 12 weeks prior to the M1 Kegworth accident (British Midlands, 1989). The aircraft had been fitted with seating that had been shown to conform with Federal Aviation Authority recommendations for static requirements in addition to improved dynamic requirements of FAR Part 25 Amendment 25-64 (AAIB 1990 and Carter 1992). These seats were designed to withstand dynamic forces of 9G in a forward direction (-Gx), 6G in a vertical direction (+Gz), and 4G in a lateral direction (Gy).

A seat plan (supplied by British Midland) of G-OBME indicated that there were twenty six rows of seats arranged six abreast (see figure 3.1.1). The seat rows were numbered from 1 to 27 with no row 13. Each row was labelled from A to F, with the aisle between seats C and D. Over-wing escape hatches were located between rows 11 and 12 and rows 12 and 14, with further escape hatches at the front and rear of the aircraft. The distance between rows of seats (pitch) was 32 inches (81.3 cm), with an increased pitch of

Seat Plan

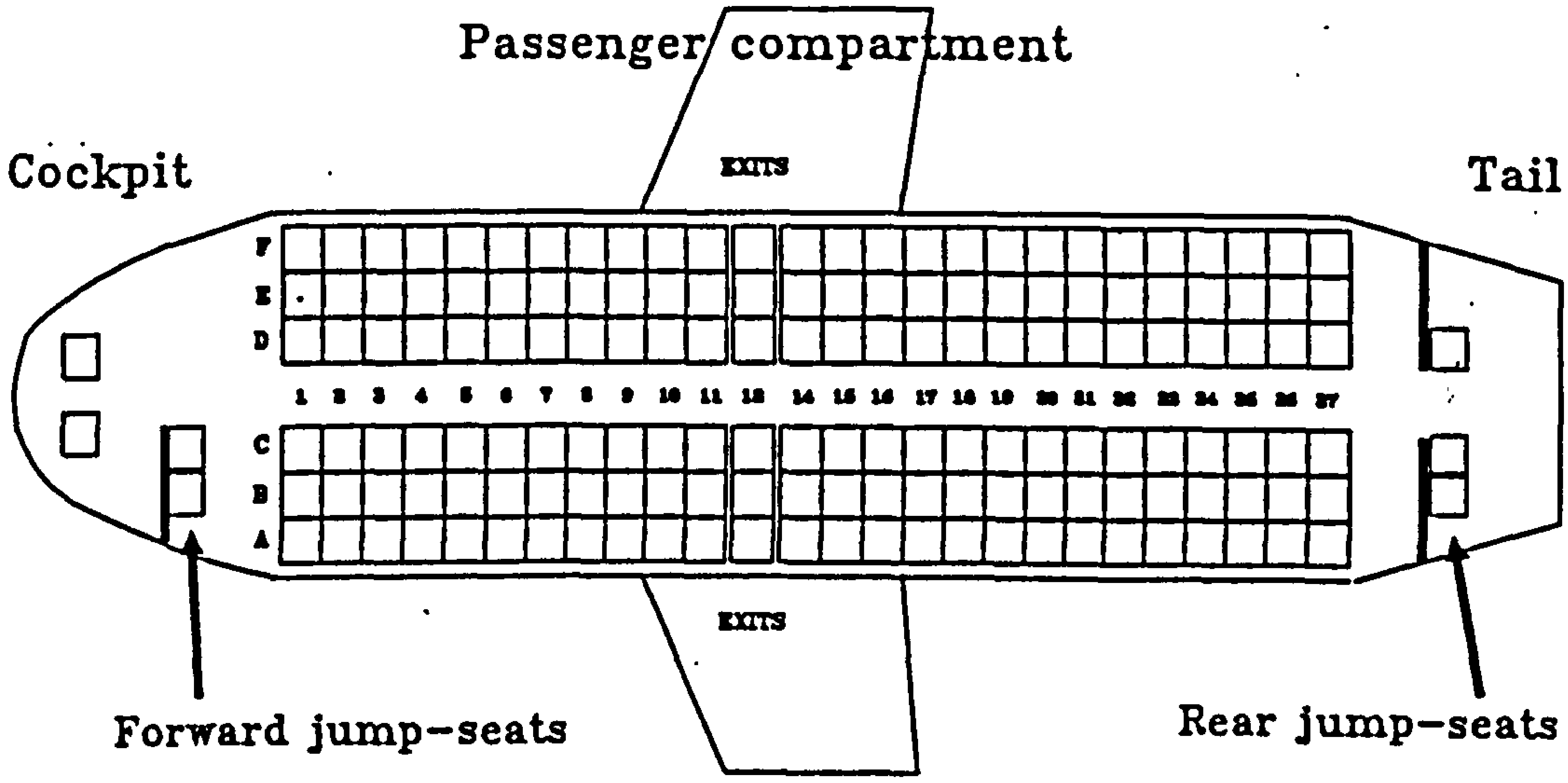


Figure 3.1.1

38 inches (96.5 cm) in those rows located around the over-wing fire escapes. Towards the rear of the aircraft the seat pitch was as small as 27 inches (68.5 cm) (AAIB 1990). The seat backs of those seats in rows 11,12 and 14 were 'locked' or fixed in an upright position, specifically to help evacuation of the aircraft, by reducing the risks of blocking the emergency exits.

Crew members were seated in rear facing seats (jump seats) located at the front and rear of the aircraft, with 2 at the front and 3 at the rear. These seats were attached to bulkheads that segregated different areas of the aircraft. The jump seats were fitted with a four point inertia reel shoulder harness as opposed to the lap type seat belts found in the passenger compartment.

3.2 The crash

On the evening of January 8 1989, an East Midland Boeing 737- 400 aircraft with only 521 hours of recorded flying time , took off on a routine scheduled flight from Heathrow to Belfast. Following vibration (later confirmed to be due to fan blade failure) in the port engine, the crew diverted the aircraft to East Midlands Airport (AAIB 1989, Costley 1989, Learmount 1990). Due to a combination of pilot error and new instrument design the undamaged starboard engine was switched off. As the final approach to East Midlands Airport was being executed, the port engine suffered a further serious mechanical failure and the aircraft crashed at 8:26pm on Sunday the 8 January 1989, whilst attempting an emergency landing. The aircraft came to rest on the western embankment of the M1 motorway on the border of three counties: Derbyshire, Leicestershire and Nottinghamshire (figure 3.2.1). A small fire in the port engine was rapidly extinguished by the airport fire tender, which arrived quickly at the scene.

The crash sequence comprised two different impacts. On final approach with reduced engine power the aircraft (in a pitch up attitude), struck the top of the eastern motorway embankment. Following this initial impact the aircraft

Figure 3.2.1 Crash Scene

(Empics, Nottingham)

rotated to a pitch down trajectory and made its final impact at the base of the western embankment on the northbound carriage way of the M1 motorway (AAIB 1989, Sadeghi et al 1989, 1991, AAIB 1990). This is illustrated in figure 3.2.2 taken from AAIB report on the accident.

Crash Sequence

(AAIB 1990)

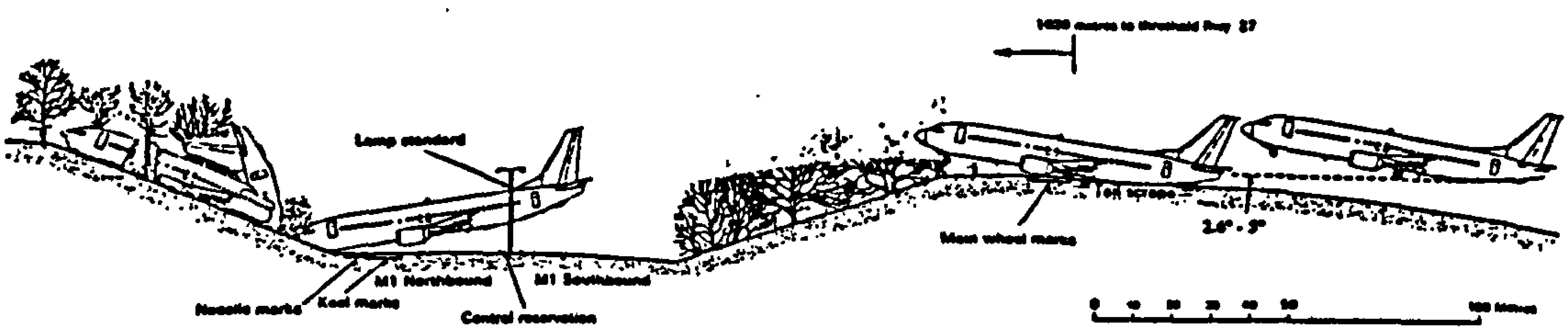


Figure 3.2.2

The fuselage of the aircraft broke into three main sections on impact. The tail section separated from the main fuselage, jack knifing over the middle section of the aircraft and crushed the roof badly over rows 18 to 22 on the starboard side and less so on the port side. The tail section came to rest having rotated through 90 degrees.

The structural damage to the aircraft's fuselage was assessed and scored according to the amount of damage sustained either to the floor, walls, or roof of the fuselage for each side, left and right. Damage was scored at each seat row on a scale of 0 to 5, with 0 the score for a normal structure and 5 indicating that a structure was absent. Thus for any given row a score of 0 indicates that the fuselage remained largely intact and a score of 30 that the fuselage was completely destroyed. Figure 3.2.3 demonstrates graphically the areas of severe damage to the aircraft.

In the central part of the aircraft the structure and seats remained relatively intact, because the centre section was part of the strong central wing torque box, an essential part of the structure of the aircraft. This region transmits forces from the undercarriage as well as the wings to the fuselage. The part of the fuselage forward of the wings sustained catastrophic failure. With the failure of the floor in this forward section the seats sustained severe damage being concertinaed into each other. Seat damage has been reviewed in AAIB report 4/90 (1990). Similarly in the section aft of the wings, the seats and structure also sustained severe damage. Seats in the last 2 rows of the aircraft remained intact.

Graphic representation of the airframe structural damage

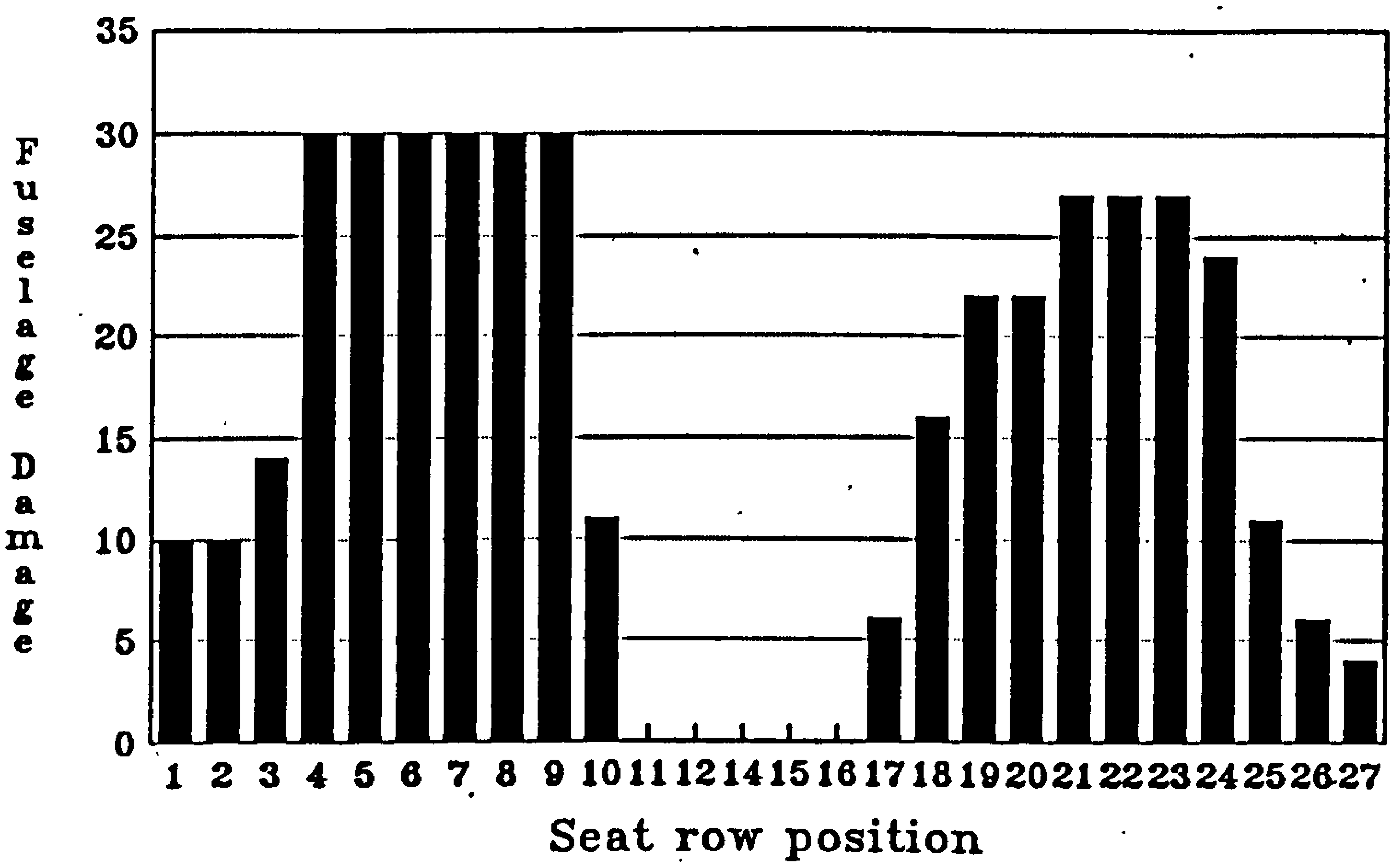
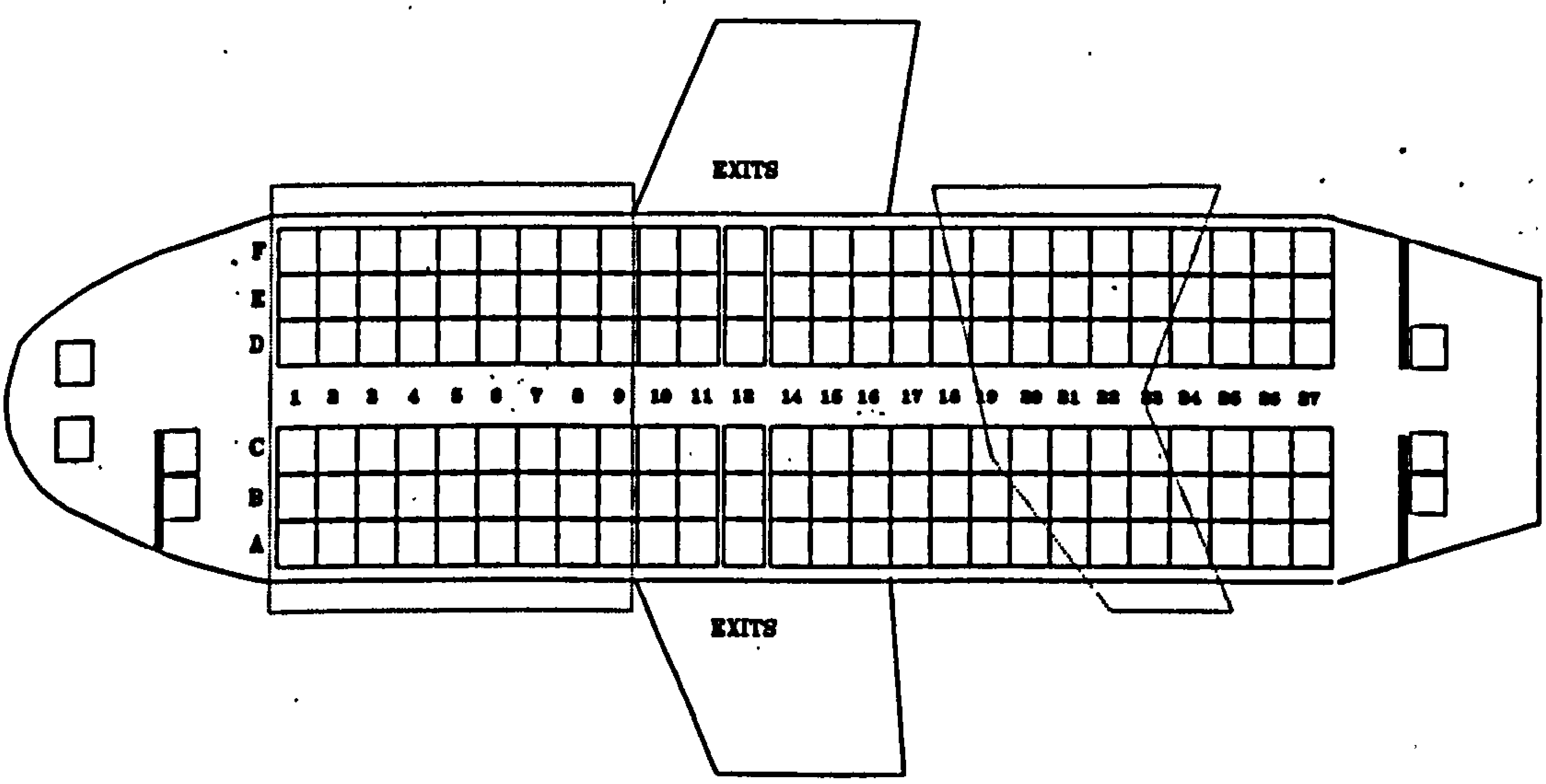


Figure 3.2.3

Areas of Structural Disruption



3.21 Accelerations Involved in the Crash

Cranfield Impact Centre Ltd. under contract to the AAIB, investigated the crash dynamics of the aircraft accident using a structural dynamics program "KRASH". This mathematical model was written initially as a research tool to define aircraft floor pulses and their effect on the seat-occupant system, and thereby general aviation aircraft crash behaviour could be modelled (Wittlin 1985). The "KRASH" program was used by Cranfield Impact Centre to investigate the overall trajectory of the aircraft as well as to predict the acceleration time-history and resultant forces likely to be experienced by the occupants during the crash (Sadeghi et al 1989). This information was subsequently used to investigate occupant dynamics by H W Structures Ltd. (H W Structures and NLDB Study Group 1990). This aspect of the collaborative NLDB work represented the first use of simulation programs to model an actual commercial jet transport aircraft accident.

The aircraft suffered an impact estimated to extend for a total period of 2.2 seconds. This began with the initial minor tail impact on the east side of the M1 motorway, until the aircraft finally came to rest on the western embankment (figure 3.2.2). The maximum acceleration calculated by Cranfield Impact Centre to have occurred at the first impact was 2.5G in the vertical direction (Gz). This initial impact was considered by the AAIB not to be significant.

Acceleration time histories for the final (second) impact on the western embankment of the M1 motorway were only considered for that part of the fuselage that remained structurally intact with large numbers of survivors; ie. the mid portion of the aircraft.

Figure 3.21.1 demonstrates the acceleration time history for the mid section of the aircraft in the horizontal (Gx) direction. It was calculated that an acceleration of approximately 15 to 20G was experienced by the occupants of the mid-section for approximately 100 milliseconds (ms).

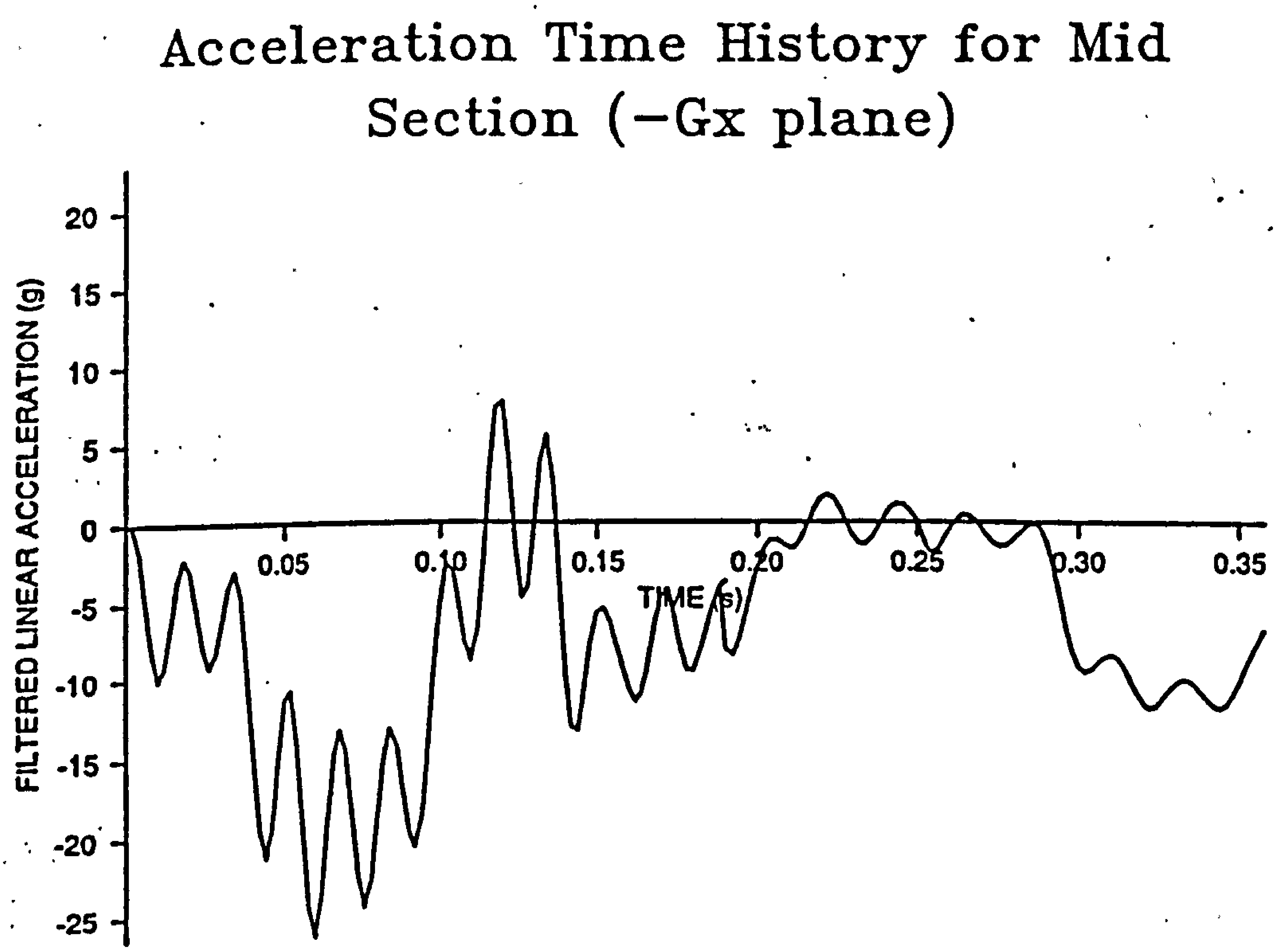


Figure 3.21.1

The acceleration time history of the mid section in the vertical (Gz) direction is represented in figure 3.21.2. This pulse occurred after the initial horizontal -Gx component and lasted for approximately 100ms. It reached a peak of some 23G (+Gz). No significant lateral (Gy) component was identified for the accident.

Figure 3.21.3 demonstrates the resultant acceleration time history for the mid section. The major accelerations were experienced in the first 200 ms of the impact sequence. The main acceleration pulse was made up of two components an initial horizontal component followed by a vertical acceleration.

3.3 Seating Plan

Essential to the analysis of any aircraft accident is an accurate seating plan of those onboard. Seating plans drawn up before a flight may be up to 30% inaccurate due to passengers moving to unoccupied seats. Mason in 1968 commented that 'The usefulness of a seating plan is often disappointing....... The Investigator is, therefore, dependent very largely on the ability of the survivors to remember the position of their neighbours'.

Using the original 'boarding' seating plan supplied by British Midland Airways, and statements from survivors a

Acceleration Time History for Mid Section (Gz plane)

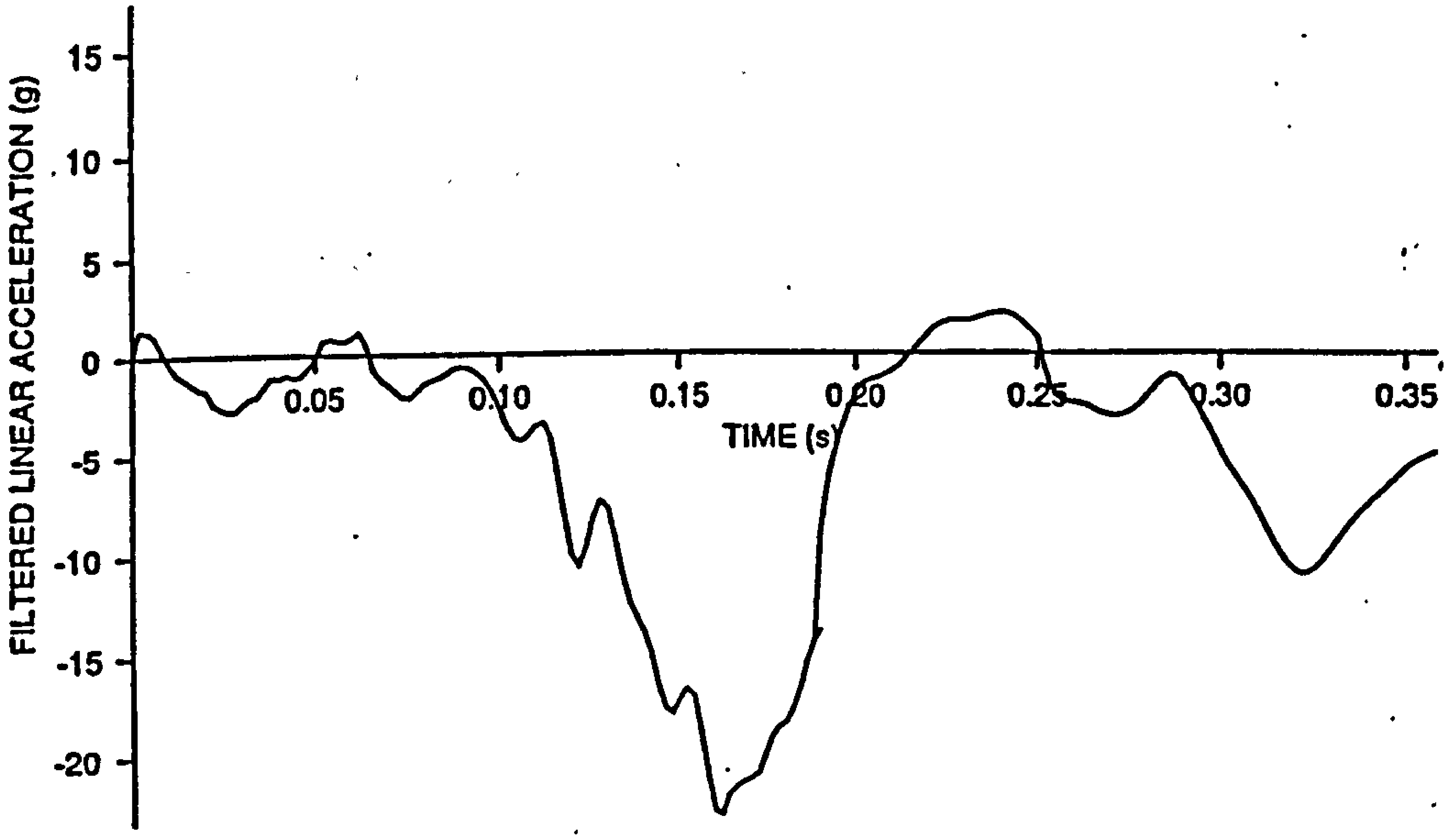


Figure 3.21.2

Resultant aceleration time history of mid setion

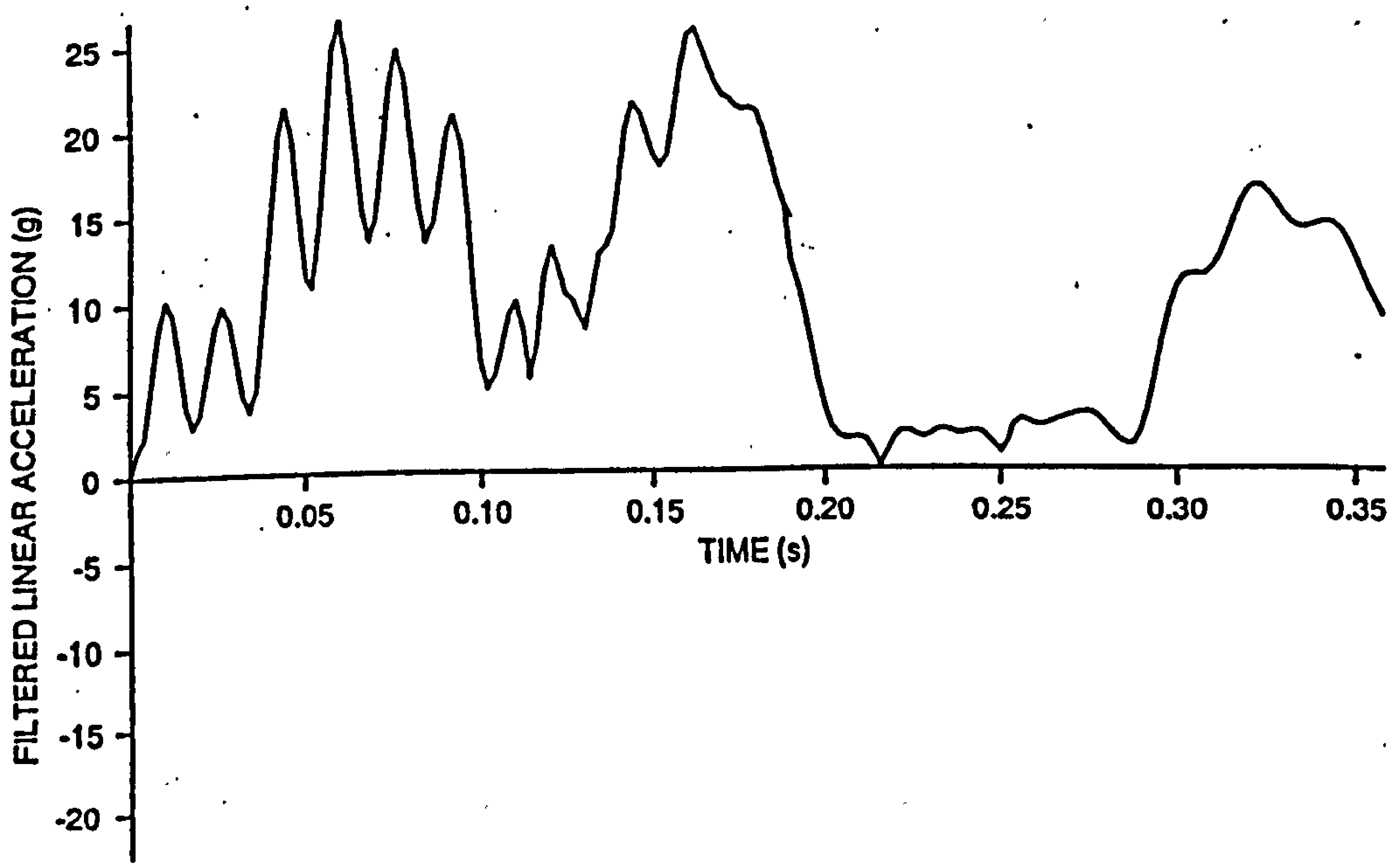


Figure 3.21.3

seating plan was constructed for occupants of the aircraft. Figure 3.3.1 demonstrates the distribution of survivors and deceased passengers on the aircraft. An accurate seat plan enables an investigator to relate the structural changes and forces involved in the crash to the injuries sustained by the occupants (Kirkham 1982).

Seating Plan

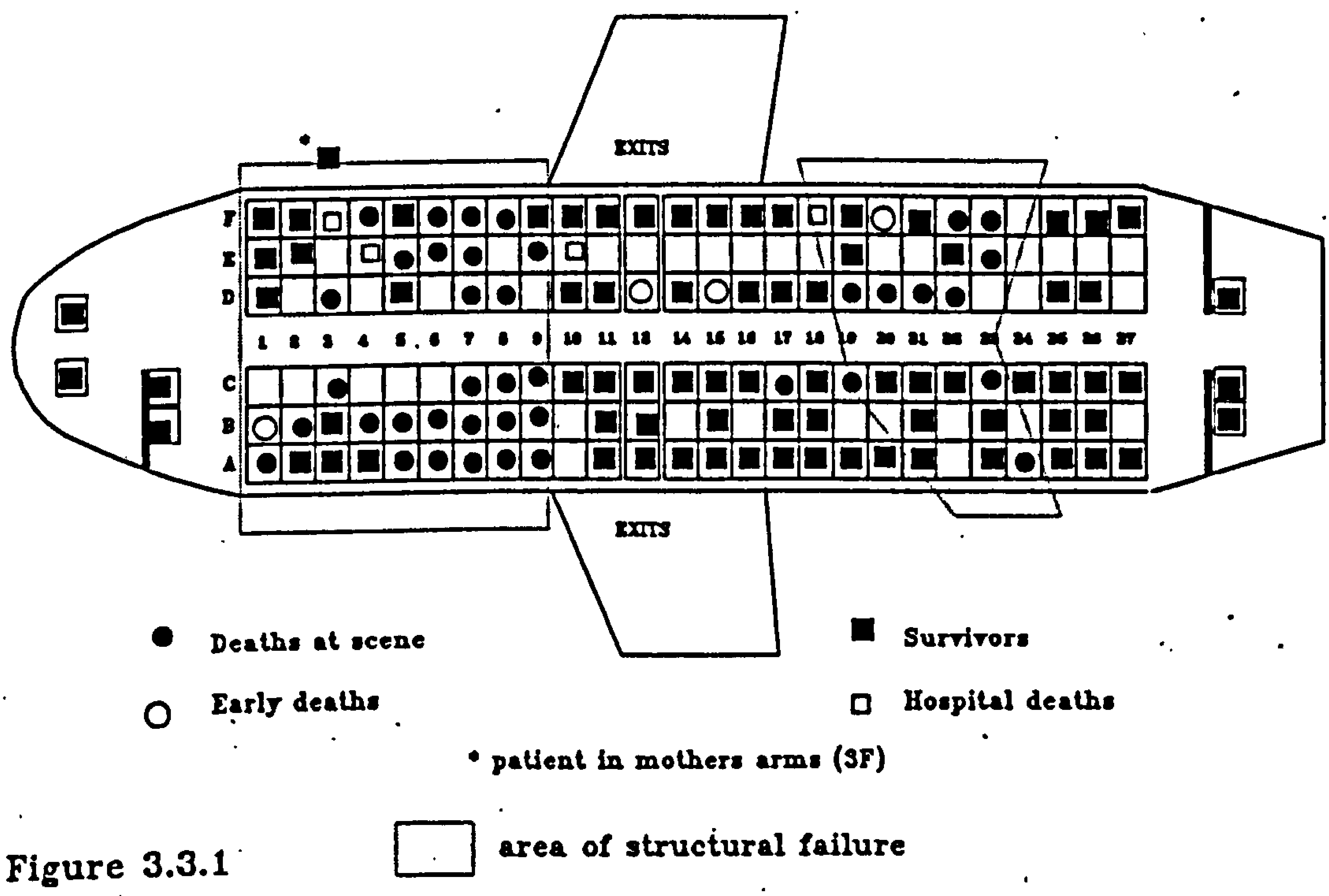


Figure 3.3.1

A survivable accident is defined as "an accident in which the forces transmitted to the occupant through the seat and restraint system do not exceed the limits of human tolerance to abrupt accelerations, and in which the structure in the occupants immediate environment remains substantially intact to the extent that a liveable volume is provided for the occupants throughout the crash sequence" (National Transportation Safety Board 1974).

The causes of mortality in an aircraft crash have been identified as; crushing within a collapsing airframe; entrapment within the wreckage; being struck by loose objects; absence or failure of restraint; injuries associated with escape; and explosive decompression (Mason 1962, Fryer 1965, Hill 1982 1984). In order to survive an aircrash a passenger requires a 'survival envelope'. The forces involved must therefore not be so great as to cause failure of the airframe, failure of the seats and restraints, and failure of cabin fitments.

Examination of the airframe of G-OBME (figure 3.2.1 and 3.2.3) revealed that severe damaged to the aircraft occurred in regions forward and aft of the wings. Because of the catastrophic failure in these regions a survival envelope did not exist and therefore one would expect a high mortality. Twenty nine of the fatalites were seated in the front section (rows 1-9), 4 were seated in the central section (rows 10-17) and 11 were seated in the rear section of the the aircraft (rows 18-27). Thus 90% of all deaths occurred in those parts of the aircraft where there was the greatest structural damage.

3.4 Injuries to Passengers and Crew

One hundred and eighteen passengers including one baby, and 8 crew members were aboard the aircraft. These consisted of 79 males (63% of all passengers and crew) and 47 females (37%). Fifty six males (71% of all males on board), and 31 females (66% of all females on board) survived the impact. A further four occupants died in hospital within a few hours and a further 4 died at variable times later in hospital. Three of the late deaths died as a result of well recognised complications of major trauma. The age distribution for all passengers and crew is demonstrated in figure 3.4.1. The majority of those onboard the aircraft were young to middle aged individuals. The injuries sustained and their survivability was reflected by this age distribution.

3.41 Injuries to Survivors

The range of injuries seen in the 87 passengers and crew surviving the crash is shown in figure 3.41.1.

Head Injuries (White et al 1990)

Seventy-seven patients had evidence of head and/or facial injuries (85%), 31 of which required treatment. Within this group 45 patients were identified as having experienced amnesia around the time of the crash. Seven patients had severe head injuries requiring neuro-surgical intervention, 5 survived but one suffered permanent severe neurological disability.

Age Distribution of Passengers and Crew

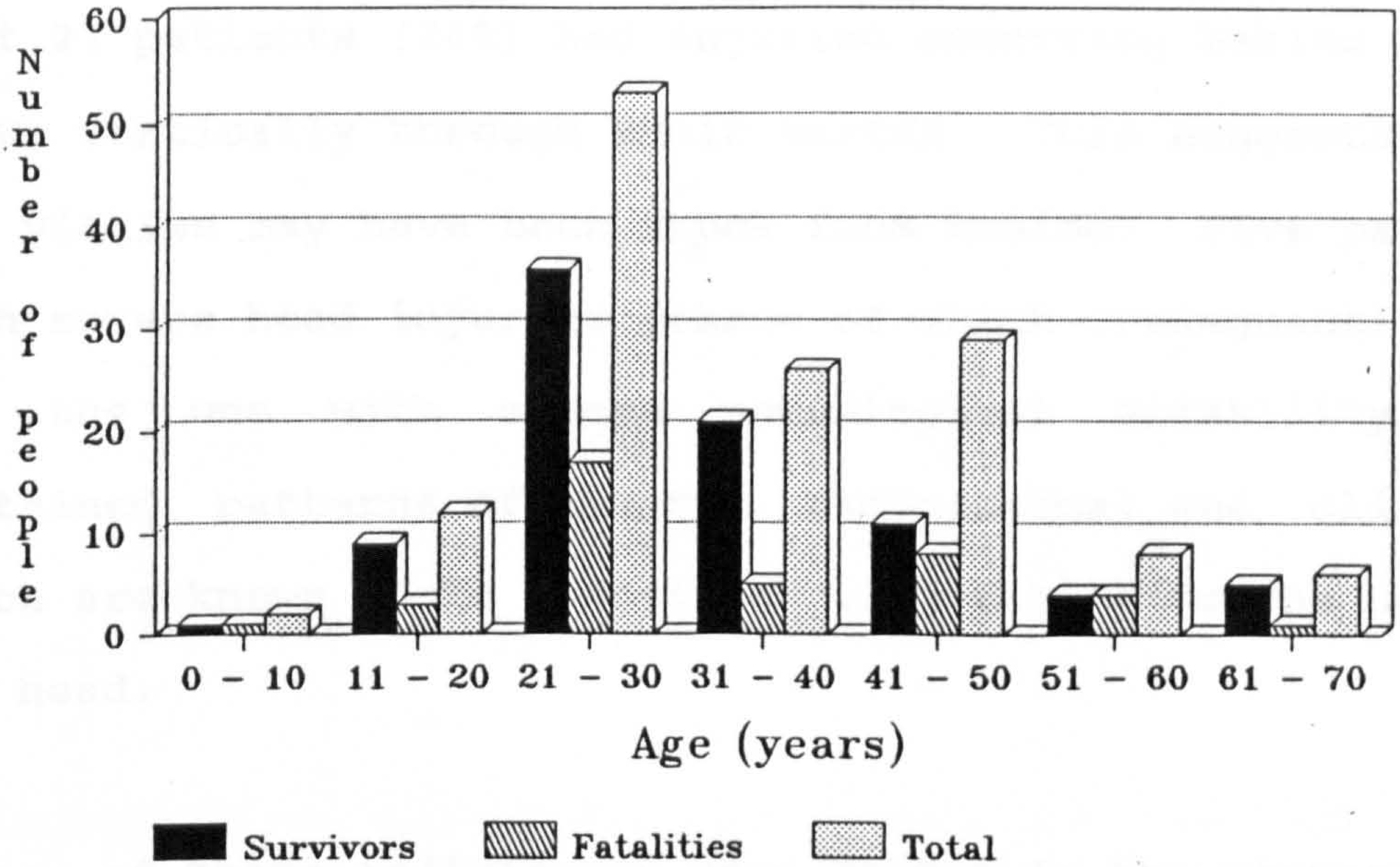


Figure 3.4.1

Major Injuries in Survivors of the
M1 Aircrash (for 87 patients surviving crash)

Region	No.
Head injury	43
Thoracic injuries	23
Abdominal injuries (operated)	2
Spinal fractures	24
Pelvic/lower limb injuries	142
open fractures	34
Upper limb injuries	59
open fractures	6

Table 3.41.1

Examination of the patterns of head injury has indicated that 21 patients (24%) had injuries occurring behind a line drawn vertically through their vertex. This suggests that the victims may have been struck from behind. Five patients with severe head injuries (three of which subsequently died and the one with severe neurological disability) had sustained patterns of injury, radiological and clinical, which are known to be associated with a blow to the back of the head.

Three of these individuals were seated in the structurally intact middle of the aircraft where high survivability was seen. It has been concluded that the cause of these head injuries was likely to be due to cabin furniture or the contents of stowage bins becoming dislodged on impact and falling on top of their heads.

<u>Chest injuries</u> (Morgan et al 1990)

Twenty-three passengers sustained major chest trauma but in addition all had injuries to other parts of their bodies. Five of these patients died within 12 hours of the accident. Fifteen patients sustained rib fractures, 11 patients had a haemo- or pneumo-thorax and 15 patients sustained lung contusions. Six patients demonstrated a widened mediastinum but investigations failed to identify a major vascular injury in any of them. Three survivors had electrocardiographic (ECG) changes consistent with myocardial contusion.

A significant correlation between the age of patient and number of rib fractures was evident (Morgan et al 1990) such that with increasing age a greater number of rib fractures were seen. Younger patients had few or no rib fractures but a high incidence of lung contusion. Pulmonary contusion in young trauma victims with no rib fractures has been well described (Locicero and Mattox 1989).

Examination of the pulmonary contusions in the chest radiographs in this group, has shown a striking "upper zone pattern" which has not previously been described. It is likely that this pattern of injury was caused as a result of the mobile thoracic and abdominal viscera being thrown forward into the rigid funnel-shaped thoracic apex. Violent compression followed by decompression forces would be concentrated on the upper lung zones.

<u>Abdominal injury</u> (Rowles et al 1990)

Despite the apparent vulnerability of the abdomen to trauma only 2 patients sustained a major intra-abdominal injury. A further 3 patients required exploratory laparotomies but these were essentially negative. Both the patients who sustained major intra-abdominal injuries, were severely injured with head, chest and limb injuries. One had sustained a ruptured bladder, the other a ruptured spleen. However 30 patients were found to have developed significant lower abdominal bruising associated with the wearing of the lap type safety belts. Evidence of such bruising following an automobile accident suggests that intra-abdominal injury should be expected (Doersch and Dozier 1967, Pederson and Jansen 1979) but this was clearly not the case in this aircraft accident. Thirteen patients had haematuria but in all cases this was managed conservatively.

There is a paucity of clinical reports on intra-abdominal injury in survivors of an aircrash, however Hill (1982) has indicated that in aircrash fatalities, abdominal injuries are usually associated with multisystem injuries.

Spinal injuries

Twenty-one passengers sustained a total of 24 spinal injuries, 6 cervical fractures and/or dislocations, 6 thoracic fractures and/or dislocations and 12 lumbar fracture and/or dislocations. Six patients suffered significant neurological injuries, 3 with a tetraparesis and 3 with a paraparesis. Spinal injuries were common in those regions of the aircraft that had sustained the most severe damage.

Pelvis and lower limb injuries

One hundred and forty-two (Abbreviated injury score {A.I.S.} >1) pelvic and lower limb injuries were identified in 57 passengers. There were 23 pelvic fracture and/or dislocations, 22 femoral fractures, 18 knee injuries (including ligamentous injuries and severe lacerations), 31 fractured tibiae, 26 ankle fractures and 22 foot injuries. Thirty four of these lower limb injuries were open.

A number of uncommon foot injuries were seen. These were 8 compound talar fractures -the so called Aviators Astragalus first described by Anderson in 1919 (Coltart 1952, Hawkins 1970), and 6 tarso-metatarsal (Lisfranc) fracture-dislocations.

Upper limb injuries

Fifty-nine upper limb and shoulder girdle fractures and/or dislocations were identified in 36 passengers. Nineteen fractures and/or dislocations were around the shoulder girdle. there were 9 humeral fractures, 21 fractures of the radius and ulna and 10 fractures or ligamentous injuries affecting the hand. Six of the upper limb fractures were open.

An analysis of the position adopted at the time of impact by each occupant has suggested an association between the type of upper limb injury sustained and the positioning of the arms at the time of impact. Bilateral or unilateral forearm fractures were sustained by some of those patients who placed their forearms horizontally in front of their faces. Whereas those patients who placed their arms vertically at the side of their heads or held on to the seat in front tended to sustain injuries around the shoulder.

3.42 Injuries to Non-survivors

Table 3.42.1 illustrates the range of injuries in the 39 passengers who died at the scene of the accident. The severe nature of many of the deceased passengers' injuries, is a consequence of their seat position in areas of the aircraft that had sustained the most severe damage.

Major Injuries in non survivors of the M1 Aircrash (for 39 scene deaths)

Region	Number
Head injury	39
Thoracic injuries	39
Abdominal injuries (operated)	31
Spinal fractures	13
Pelvic/lower limb injuries	95
Upper limb injuries	22

Table 3.42.1

Head, chest and abdomen

All the thirty nine passengers who died at the accident scene sustained injuries to their heads ranging from lacerations to severe disruptions. All non-survivors were found to have had injuries to the chest wall or to their thoracic viscera. Head and chest injury were the leading cause of mortality. This is not surprising taking into account the severe damage sustained by the aircraft in those regions where the mortality was high. Head and chest injuries have been recognised previously as being the leading factors in mortality from aviation accidents (Swearingen et al 1962, Mason 1962 1973, Gilles 1965, Steven 1970, Hill 1984).

In contrast to those who survived the majority of on site fatalities had sustained intra-abdominal injuries. This is in keeping with Hill's (1982) observation that there is a high incidence of hepato-splenic injuries in fatalities of aviation accidents and these are usually associated with severe multisystem injuries.

Spinal injuries

The spinal injuries identified in the deceased were all of a major degree (AIS > 4). One lumbar spinal fracture was recorded, 6 thoracic fractures and/or dislocations and 6 cervical fractures and/or dislocations. Thoracic fractures and/or dislocations were often associated with an aortic transection and a sternal fracture.

It is likely that some spinal fractures were not identified at necroscopy in the non-survivors. Routine radiology of the skeletal system of the deceased was not carried out but had it been it is likely that further spinal injuries and possibly other fractures could have been revealed. Hill (1984) has commented that the inadequacy of clinical and autopsy reports often leads to a lack of realism in planning and interpretation of experimental studies of injury production. Radiology has been used extensively for identification purposes (Hill 1979, 1984, Lichtenstein et al 1980) and as an investigative tool in aviation accidents in the armed services (Lichtenstein et al 1980). The use of post-mortem X-ray studies in civilian air accidents is limited. Lichtenstein has explained how known mechanisms of trauma often produce distinctive injury patterns. The use of radiology has probably been underutilised in the past as it may be used to deduce the mechanism of injury and the crash kinematics and may help in crash reconstruction.

Pelvic and limb injuries

Upper limb and shoulder injuries occurred in 18 of the fatalities with a total of 22 injuries identified. Humeral fractures were the commonest with 9 fractures seen. Seven wrist fractures and/or dislocations were also recorded with the remainder of injuries distributed between the hand, forearm, elbow and clavicle. One victim sustained a traumatic amputation of an upper limb. Seven of the upper limb injuries were open (compound) fractures.

Thirty five of the 39 non survivors sustained a total of 95 pelvic and lower limb injuries (AIS > 2). Only four of those who died at the scene did not sustain a lower limb injury. Nine pelvic injuries were seen, 13 femoral fractures, 5 knee injuries, 38 lower leg fractures, 24 ankle fractures and 6 foot injuries. Forty nine of these injuries were open (compound) fractures (50%).

3.5 Injury Scoring in the Evaluation of the Injuries Sustained (Rowles et al 1992)

With the exception of the abdominal injuries there is little to distinguish the actual types and severity of the injuries sustained by the non survivors compared with the survivors. Variations in the injuries will be considered further using injury scoring techniques.

3.51 The Abbreviated Injury Score (AIS)

Injury scoring as a means of classifying the extent of trauma has a long history. The abbreviated injury score (AIS) has been used extensively for assessing the severity of road related impact injury (Aldman and Chapon 1984, Viano and Levine 1986). The AIS is a threat to life score and has become the universally preferred system for assessing the severity of impact injuries (Petrucelli 1984). However the majority of impact injury patients die because of more than one injury. Injuries that in themselves would not be life threatening could have a significant effect on mortality when combined with other injuries. A scoring system that can assess multiple injured patients has thus been developed, by Baker et al (1974), the 'Injury Severity Score' or ISS.

For the AIS the body is divided into six regions as described by the American Association for Automotive Medicine Abbreviated Injury Score (AIS 1985 revision) - head and neck, face, chest, abdominal and pelvic contents, extremities and pelvic girdle, and general (external). The injuries are scored in each region with an increasing severity from 1 - 5. A score of 1 is considered a minor injury whereas a score of 5 is considered a critical injury where survival is unlikely. A score of 6 is possible in any given region but is considered to be non-survivable (in AIS 85).

The regional AIS's were calculated for every person on the aircraft using the clinical notes and/or post-mortem findings. Table 3.51.1 shows that a total of 432 injuries were identified in the 83 passengers and crew who survived the impact and were admitted to hospital. The majority of injuries occurred in relation to the limbs and pelvis, and soft tissues.

Table 3.51.2 shows the average abbreviated injury scores for for each of the body regions for both non survivors and survivors. The severest injuries were to the head and chest

Table 3.51.1

Regional Abbreviated Injury Scores(AIS)

(N = 83 PATIENTS)

Body Region	AIS 1	2	3	4	5	TOTAL
Head or neck	5	15	14	6	3	43
Face	4	2	-	1	-	7
Chest	6	6	12	6	1	31
Abdominal / pelvic contents	11	15	5	3	1	35
Extremities / pelvis	20	117	68	-	-	205
External	104	6	1	-	-	111
TOTAL	150	161	100	16	5	432

Table 3.51.2

Average AIS in Occupants

	Average AIS Survivors	Non Survivors	Significance p=<
Head and neck	1.2	3.7	0.0005
Face	0.2	1	0.002
Chest	1.1	4.6	0.0005
Abdomen and pelvic contents	0.7	2.5	0.0005
Limbs and pelvic girdle	2.1	2.8	0.0002
External	1.2	1.7	0.0002

regions in non survivors. Significant differences in the severity of the regional injuries between survivors and non survivors exist.

3.52 Injury Severity Score (ISS)

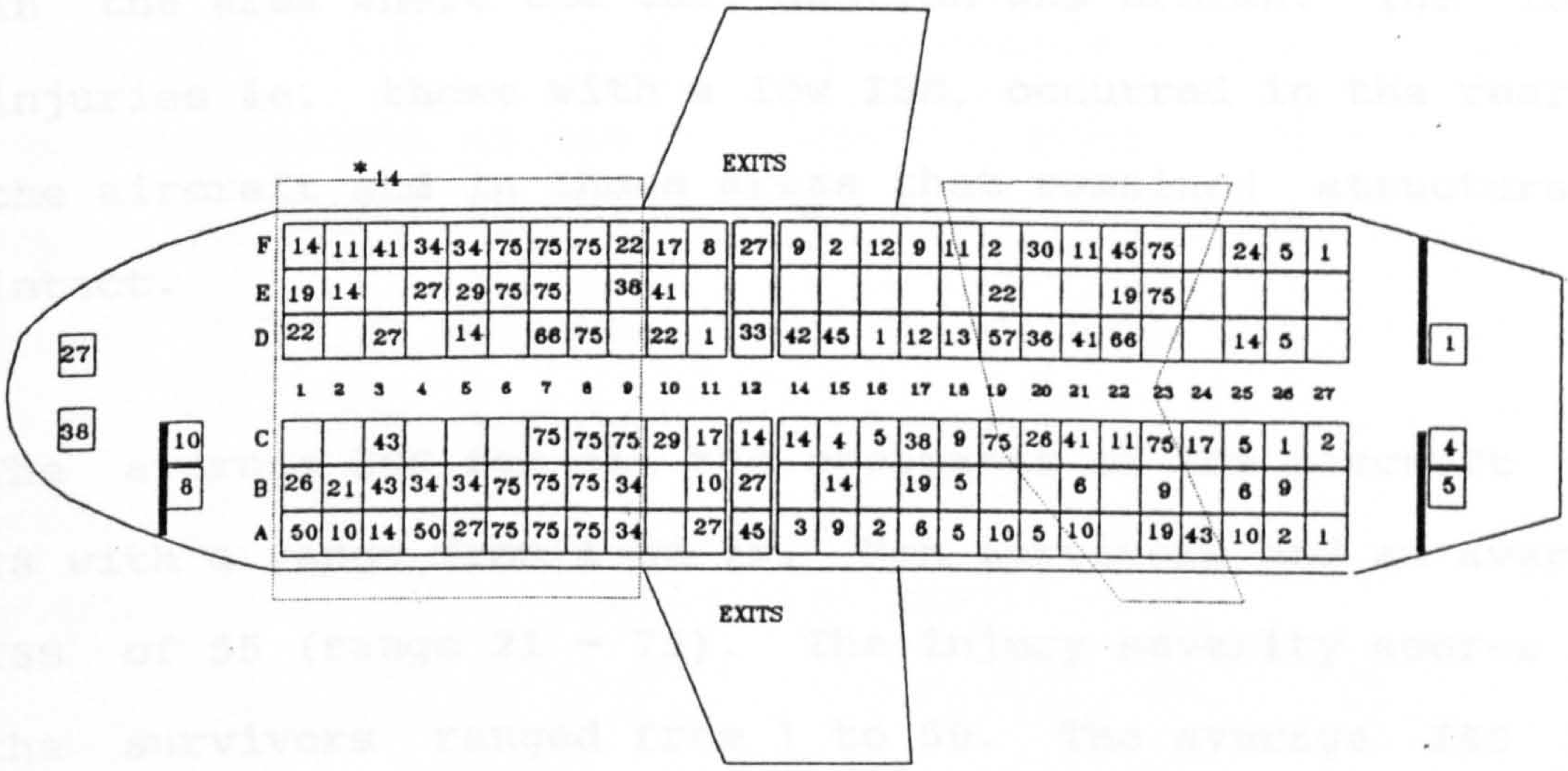
Injury severity scoring has become the established scoring system for survival prediction and trauma audit. It is an index of anatomical injury, but takes no account of the physiological or psychological effects of trauma. The score has been found to be useful as a measure of mortality, survival time, hospital length of stay and disability (Bull 1975). This scoring system has led to definition of a patient with "major trauma", that is a patient who scores an ISS of sixteen or more points. An ISS of 16 is predicative of a 10% mortality (Boyd et al 1987).

The Injury Severity Score is calculated as the sum of the squares of the 3 highest regional AIS (Baker et al 1974). Therefore $ISS = AIS_{(1)}^2 + AIS_{(2)}^2 + AIS_{(3)}^2$, where $AIS_{(1)}$, $AIS_{(2)}$ and $AIS_{(3)}$ are the three most severely injured regions. Only the contribution of the most major regional injuries is used. The highest ISS attainable is 75 (ie. $5^2 + 5^2 + 5^2$) or if any body region scores an AIS of 6, an ISS of 75 is automatically scored.

The Injury Severity Score is thus a system for denoting the magnitude of injury on a non linear discontinuous scale from 0 to 75 (Stoner et al 1977). Zero represents no injury whereas 75 denotes an injury that is considered non-survivable.

Figure 3.52.1 shows the ISS of each occupant according to his seat position.

ISS for Passengers and Crew



* patient in mothers arms (3F)

Figure 3.52.1

The structural damage to the aircraft's fuselage has been discussed in section 3.2. The structural damage sustained by the aircraft for each seat row was compared with the ISS's of the occupants in that row. As can be seen the ISS correlated well with those regions of the aircraft which had sustained the most severe damage (Spearman rank correlation, rho = 0.569 with 116 d.o.f. p=<0.0005). Thus a high ISS is associated with a high aircraft structural damage score.

The highest scores (the most severe injuries) occurred in rows 6 to 8 in the region forward of the fuselage break. Serious injuries also occurred in the whole area forward of the wing. Further serious and fatal injuries occurred in the region of the failure of the rear fuselage and floor, in the area where the tail section was broken. The least injuries ie. those with a low ISS, occurred in the rear of the aircraft and in those areas that remained structurally intact.

The average ISS for all the occupants of the aircraft was 28 with a range from 1 to 75. Non survivors had an average ISS of 55 (range 21 - 75). The injury severity scores for the survivors ranged from 1 to 50. The average ISS for those 87 survivors of the impact was 16. Surviving passengers seated in the mid section of the aircraft (rows 10-20) demonstrated ISS's from 1 to 45. The wide variation indicating the complex multifactorial nature of injury causation in impact aircraft accidents.

Thirteen patients with minor injuries indicated low ISS's (less than 3) had soft tissue injuries or minor fractures. Thus 74 passengers (85%) sustained significant trauma.

The attempt by some passengers to protect adjacent passengers resulted in a higher than expected ISS in the "protector". For example (refer to figure 3.52.1) the patient seated in 3B protected the passenger in 3A by putting an arm across the shoulders of 3A forcing 3A to adopt a brace position. Injury severity score indicates that 3B (ISS 43) was significantly more injured than 3A (ISS 14). The passenger in 3F whilst protecting an infant sustained a ISS of 41 whilst passengers seated around her had lower scores as did the infant. Similar findings were found in those individuals seated in Row 9 seat numbers E and F, where 9E (ISS 41) protected 9F (ISS 22). In the rear of the aircraft in row 26 the passenger in seat 26B (ISS 9) protected the passengers in 26A (ISS 2) and 26C (ISS 1) by placing their arms across their shoulders.

By scoring passengers injuries in this way variations in the severity of injuries amongst passengers and crew in the same region of the aircraft have been identified. This information indicates that factors other than the force of the impact and individual variations may be involved in injury causation.

3.6 Managing the aftermath : Initial response of emergency services

Within minutes of the accident occurring emergency services from the three counties involved in the accident were mobilised, medical teams were dispatched, and major incident plans were put into operation by the participating medical centres. The involvement of three regions had the advantage that adequate health services were available but was associated with the disadvantage of complex organisational and communication problems.

The injured were taken to one of four hospitals in the region, University Hospital Nottingham, Leicester Royal Infirmary, Derby Royal infirmary and Mansfield General Hospital (figure 3.6.1). Four passengers died soon after

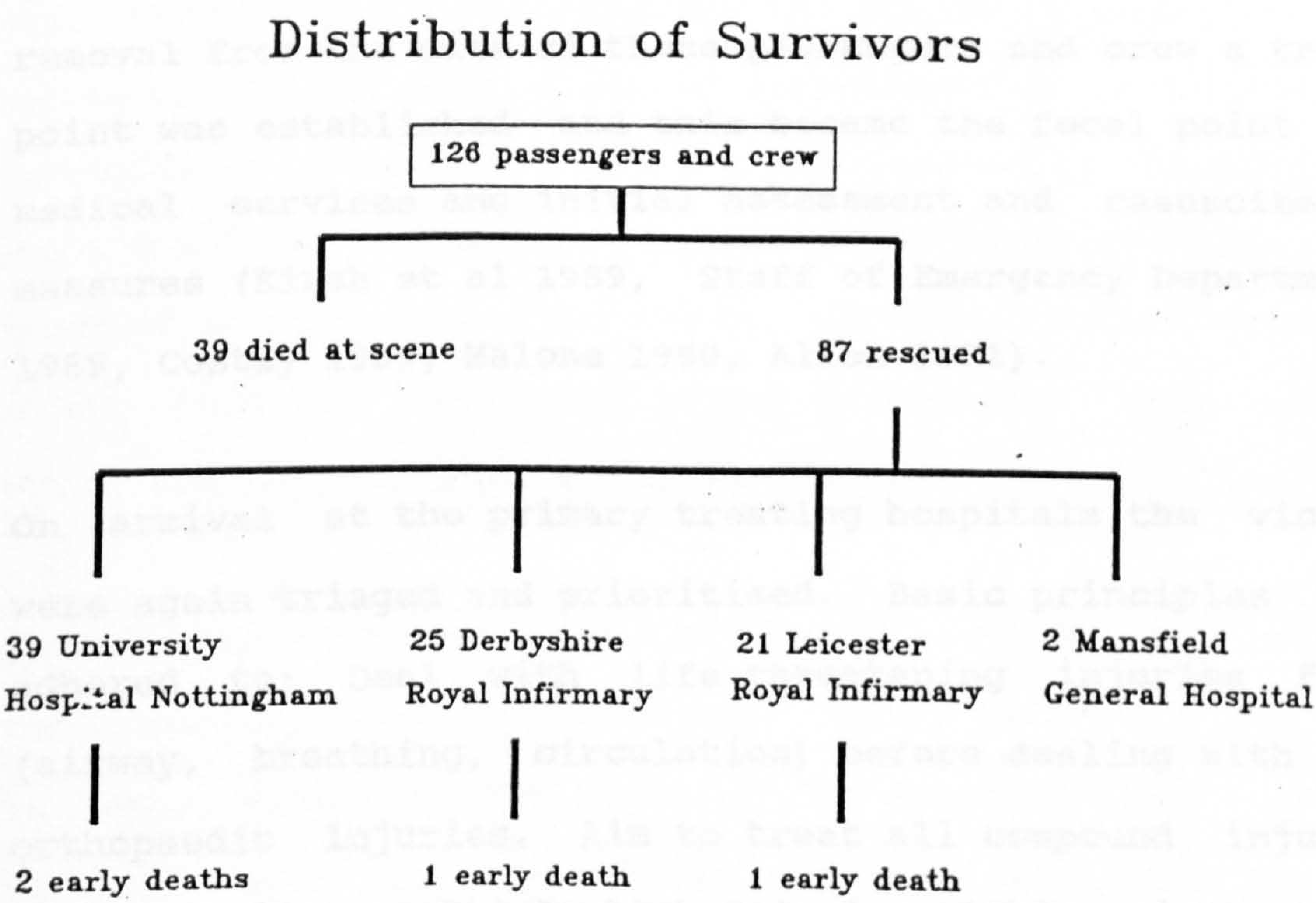


Figure 3.6.1

arrival at the primary treating hospitals. A further 4 patients died in hospital (Kirsh et al 1989). Seventy nine passengers and crew were still alive 2 years later (8 January 1992).

Within forty-five minutes of the accident the first casualties were arriving at the primary hospitals. The initial 30 casualties to be evacuated from the site were bundled on the first ambulances at the scene on a scoop and run basis. This meant that only very simple triage had been carried out on these individuals and because of poor communications they were nearly all sent to the same hospital with little warning. However after the initial removal from the site of these passengers and crew a triage point was established and this became the focal point for medical services and initial assessment and resuscitative measures (Kirsh et al 1989, Staff of Emergency Departments 1989, Costly 1989, Malone 1990, Allen 1991).

On arrival at the primary treating hospitals the victims were again triaged and prioritised. Basic principles were adhered to: Deal with life threatening injuries first (airway, breathing, circulation) before dealing with the orthopaedic injuries. Aim to treat all compound injuries and stabilise multiple limb injuries within eight hours then carry out definitive treatment of other injuries later

(Wallace 1989).

Because the accident happened on a "quiet" Sunday evening resources proved adequate. Blood stocks for the forthcoming week of surgery were available. Beds were vacant awaiting admissions on the following Monday and the Intensive therapy units were quiet as were the operating theatres.

Staffing - medical, nursing and ancillary was not a problem. "We heard it on the news so we came in " was a common response. The only problem was there were so many doctors and other staff that it was often difficult to co-ordinate their efforts. Some were sent home, their energy and efforts being reserved for the following days. On the night of the incident eighteen orthopaedic surgeons and six consultants from the accident and emergency departments were involved in treating the victims. Their numbers were matched by consultants in general surgery, anaesthesia and other surgical specialities and in addition many middle grade trainees offered their services.

On the night of the incident fourteen operating theatres were opened: eight in Nottingham, two in Leicester and four in Derby. Within twelve hours of the accident (table 3.6.2) 94 operations had been carried out on 34 survivors in the main operating theatres. By thirty-six hours after the accident this figure had increased to 136 procedures on 55 survivors with some patients receiving second operations the following day.

Twenty-two patients required admission to the intensive therapy units on the night (table 3.6.2). Later one patient required specialised treatment after the development of a fat embolism syndrome on the third day.

The radiological departments provided a speedy and efficient service. Within 12hrs of the accident 772 radiological examinations had been performed (table 3.6.2), including 6 CT scans of the head, one CT scan of the thorax and 6 angiograms (Kirsh 1989, McConachie et al 1990). Eventually over 1500 radiological investigations would have been performed on the survivors of the accident in the primary treating hospitals.

Transfusion requirements were heavy (table 3.6.3). Within 2.5 hours of the accident occurring 209 units had been cross matched. No patient received uncrossmatched blood whilst in hospital. In total 751 units were crossmatched with 596 units of blood being used. Ninety-one units of platelets and 59 units of fresh frozen plasma were also transfused (Kirsch et al 1989). Ten patients required transfusion of more than 20 units of blood with four individuals requiring more than 50 units.

Table 3.6.2

Summary of data from hospitals dealing with survivors

	Univesity Hospital Nottingham	Leicester Royal Infirmary	Derbyshire Royal Infirmary	Mansfield General Hospital
No. of patients recieved	39	21	25	2
No. of operations:				
On first day	38	23	31	2
On second day	26	-	18	-
No. of patients having operations:				
On first day	11	10	11	2
On second day	11	-	10	-
No. of patients admitted to ITU	12*	6	4	-
No. of radiographs within 12 hours	409	210	137	16

* one patient admitted later with fat embolism

Table 3.6.3

Transfusion requirements

	No. units crossmatched	No. units transfused
Whole blood		
First 2.5 hrs.	209	-
Within 12 hrs	-	382
Total used in 36hrs.	751	596
Fresh frozen plasma	-	59
Platelets		91

3.61 Primary hospital Management

Of the eighty seven passengers surviving the impact 83 passengers and crew survived to be admitted to a hospital ward. Four patients died within ninety minutes of their arrival in hospital. Two from head injuries, and the other two from multiple injuries. Three patients died during their primary hospital stay: One at 12 hours from a head injury (ISS 41). A second died 11 days after admission with multiple system organ failure (ISS 27) and a third at 15 days following a pulmonary embolism (ISS 41). A further patient (ISS 11) died on transfer to another hospital at 22 days following the accident also from a pulmonary embolism.

Only three patients were discharged from hospital on the night of the accident. The majority (58 patients or 70%) of those admitted were discharged from the primary treating hospitals within three weeks (figure 3.61.1). However a

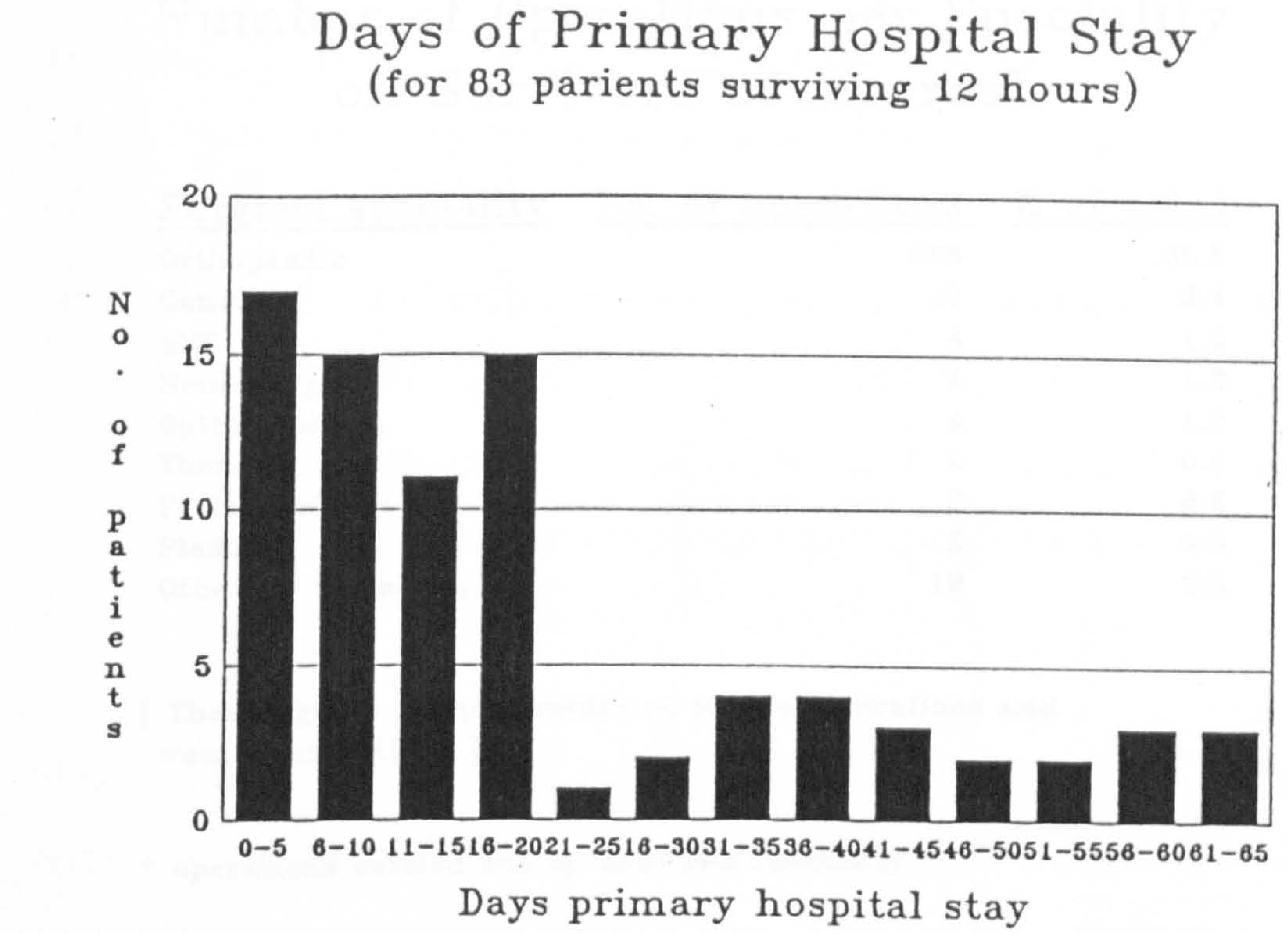


Figure 3.61.1

significant number remained after this time with some staying in hospital for more than two months (Rowles 1990).

The work load of the different surgical specialities is demonstrated in table 3.61.2. In total 345 operations were carried out on the 83 patients. The majority of procedures were orthopaedic related. Initial orthopaedic management (ie. within the first 24 hrs) involved the resuscitation of patients, surgical treatment of open fractures and reduction and stabilisation of fractures and dislocations. In the days following the accident the surgical procedures included second look procedures for open wounds and further debridements, split skin grafting, and fracture fixation

Number of Operations per Speciality on Survivors of Aircrash

Surgical speciality	No. of operations	% of total
Orthopaedic	295	85.5
General	13	3.8
ENT	4	1.2
Neurosurgery	4	1.2
Opthalmology	4	1.2
Thoracic	2	0.6
Faciomaxillary	2	0.6
Plastic	2	0.6
Others (* unknown)	19	5.5

[These figures exclude suture of simple lacerations and wound inspections]

* operations carried out by unknown speciality

Table 3.61.2

(159 procedures). Forty nine patients had orthopaedic fixation with metalwork (screws, plates, wires and nails) and one patient required a shoulder hemiarthroplasty.

The normal working practice of the primary hospitals was disrupted for many weeks following the accident. Routine orthopaedic emergency work still had to be carried out but elective admissions were cancelled because of the high bed occupancy. Indeed elective surgery requiring intensive care facilities were cancelled for three weeks at University Hospital Nottingham.

3.62 Secondary Hospital Management

Review of the survivors of the aircrash has revealed that at the time of discharge from the primary hospital, 36 (45%) of the 80 survivors were transferred to further hospitals for treatment and rehabilitation (Rowles 1990). One patient unfortunately died in a secondary treating hospital. Three patients still remained in hospital one year after the crash. At nine months after the accident 49 patients (62%) still required out patient review and 27 (34%) had been discharged.

The third phase of management, undertaken mainly in secondary hospitals involved the rehabilitation of the patients, and operations to treat the complications and the Delayed Diagnosed Injuries (Tait et al 1990). By October 1989 21 patients had had a further 49 operations (table 3.62.1) performed for the treatment of their injuries in

Operative Procedures in Secondary Hospital

Procedure	Number performed
Removal of metal work	16
Late debridement of wounds	9
Fracture fixation	9
Bone graft	7
Skin graft/plastic procedures	5
Others	3

Table 3.62.1

secondary hospitals, with 24 of these operations required for 4 patients. These operations included orthopaedic treatments for non unions and delayed unions, late fracture fixations, removal of metal and in one case the removal of a prolapsed thoracic disc. Plastic surgical treatments were required for the treatment of tissue loss, and soft tissue injuries (Rowles et al 1991).

3.7 The Consequences of the Injuries

Early complications

Four patients developed a clinically diagnosed pulmonary embolism, of which two died, and one patient was identified as having a deep vein thrombosis whilst in a primary treating hospital. Only thirteen patients received prophylactic heparin treatment for venous thrombosis. Five cases of septicaemia were recorded. A further patient developed septicaemia and pulmonary embolism four months after the accident. Two patients developed fat embolism syndrome, one requiring ventilatory assistance.

Of the 28 patients sustaining 40 lower limb open fractures there was a 30 % incidence of wound problems (Learmonth et al 1991). One case of sepsis resulted in the amputation of a leg. No cases of late sepsis occurred after nine months.

The nursing problems resulting from the management of the multiple injured patients was highlighted by the 6 patients who developed pressure sores.

Residual morbidity

Despite strong efforts to provide the optimal treatment for these severe injuries the end results have in some cases been less than perfect. Table 3.7.1 illustrates the outcome of the injuries sustained in the survivors of the aircrash after nine months. Forty six passengers (58%) have made good recoveries, 25 (32%) are moderately disabled and 7 (9%) are severely disabled. Of these seven, five had sustained spinal injuries with neurological loss and one patient has a major neurological disability as a result of a severe head injury. These six patients represent the most

Outcome of Injuries
(for 79 survivors at 9 months)

Injury severity score	Good recovery	Moderate disability	Severe disability	Vegetative
0-8	26	3	-	-
9-15	13	10	1	-
16-24	3	10	-	-
25-32	4	1	1	-
33-40	-	1	2	-
41-48	-	-	2	1
49›	-	-	1	-
Total	46	25	7	1
% of total	58%	32%	9%	1%

Table 3.7.1

severely injured passengers on board the aircraft. Their average ISS was 41 (range 32-50).

The late impairments demonstrated by the other survivors relate to complications arising from their fractures, and immobilisation which resulted in joint stiffness, wasting of muscles, limp, deformity and pain. A joint injury may have initially done well but as time progresses the joint may deteriorate and painful secondary arthritis ensue (Levine 1986). The future impairments and disabilities that will become manifest with time are likely to continue to require further hospital treatment.

Return to work

Orthopaedic injuries have been cited as the most frequent cause of serious disability following trauma. Lower limb injuries in particular are second only to head injury in causing permanent impairment and disability (States 1986). Extremity injuries on there own are rarely fatal but often require significantly longer periods of hospitalisation and more lost working days than injuries to other body regions at the same AIS level (Nyquist and King 1985). Orthopaedic related injuries usually take months to heal and even with optimal treatments a significant percentage of those injured will have a permanent impairment.

Seventy of the surviving 79 passengers and crew, were in employment prior to the crash. Forty two (60%) patients had returned to their previous occupation nine months after the accident, either full time or part time (table 3.7.2). Their average ISS was 11. The 28 individuals who had not returned to work by nine months had an average ISS of 22. The nine survivors not at work prior to the crash included five housewives, two pensioners, one person who was unemployed and one baby.

MacKenzie (1986) in his review on lower limb trauma in automobile accidents in America found that 56% of individuals who were employed prior to injury had returned to work within one year. Our figures are comparable to this study.

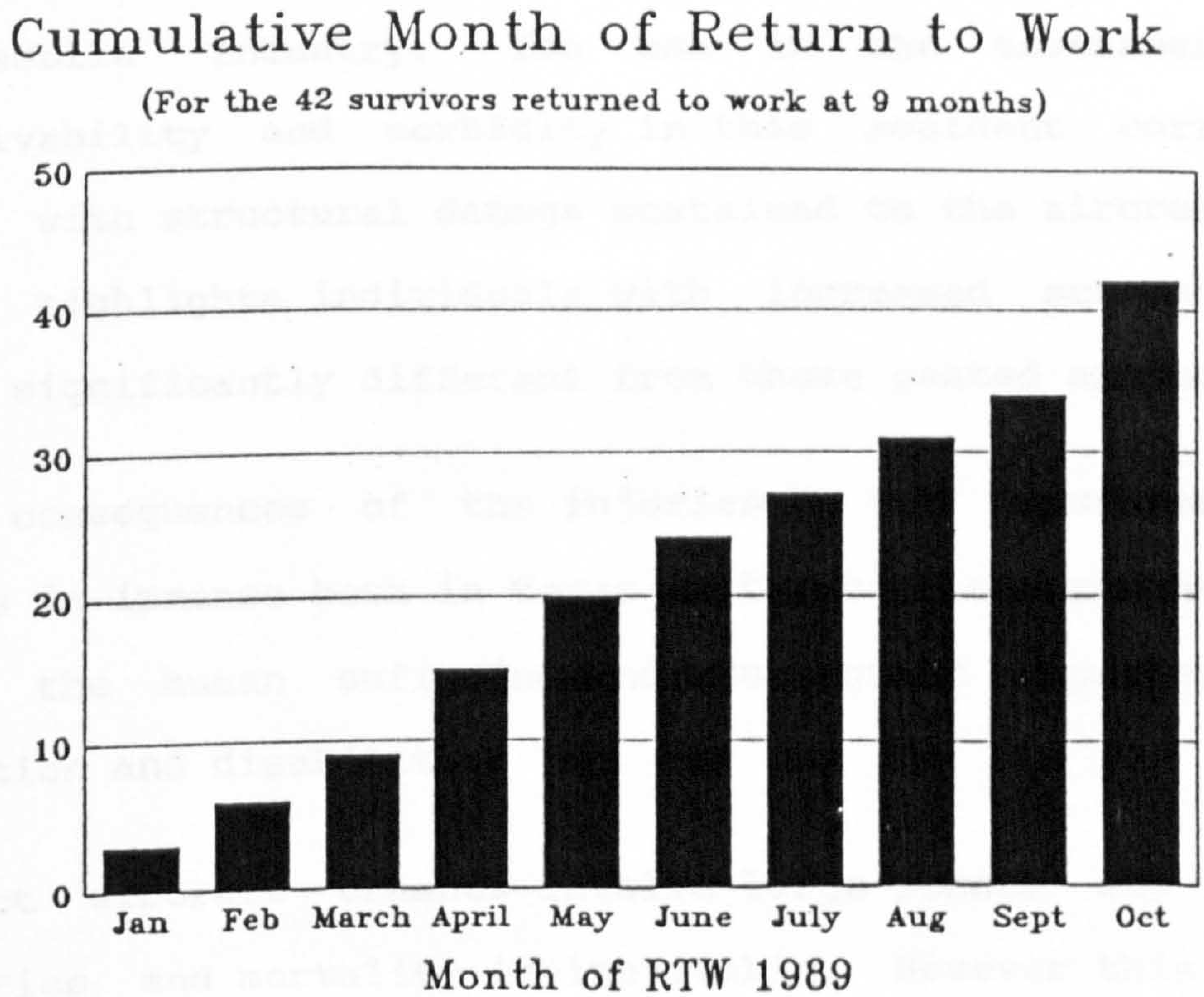


Figure 3.7.2

It is not possible to calculate the cost of treating the survivors of the M1 aircrash. In addition to the costs of hospitalisation, there are the costs of other services, such as therapists, equipment and appliances and modifications to domestic housing. In addition there is the cost to the individual of lost earnings, the psychological morbidity and social incapacitation.

3.8 Conclusions

The 87 initial survivors of the M1 aircrash sustained a variety of injuries from bruising to multiple life threatening injuries. Injury scoring has proved a useful tool in assessing the severity of impact injuries in the automobile industry. Its use in the assessment of survivability and morbidity in this accident correlates well with structural damage sustained to the aircraft. It also highlights individuals with increased scores which were significantly different from those seated around them.

The consequences of the injuries to the passengers and crews is immense both in terms of the cost of treatment and also the human suffering and subsequent impairment of function and disability.

Impact aircraft crashes involve large forces and severe injuries and mortality is inevitable. However this crash has demonstrated that it is possible for large numbers of occupants to survive. Despite the structural integrity of the centre section of the aircraft and the maintenance of the restraint systems, the passengers in this centre section sustained some severe injuries.

Future designers of aircraft must not only concentrate on reducing the number of fatal injuries, but also examine the causation of the non-fatal long-term disabling injuries in an effort to reduce their incidence. Finally it should be noted that only if the victim of an aircrash remains relatively injury free can he escape from an aircraft in the event of a fire.

Chapter 4

An Overview of the Pelvic and Lower Limb Injuries in the Passengers and Crew

4.1 Introduction

This chapter reviews the incidence and types of pelvic and lower limb injuries sustained by the passengers and crew, on board Boeing 737, G-OBME, when it crashed onto the M1 motorway. The methodology is described in the first part of the chapter followed by an overall impression of the pelvic and lower limb injuries in all the occupants. The second half of the chapter then considers the detailed biomechanical features of injuries in those survivors seated in the centre section of the aircraft. In the mid section the seating and restraint system remained, on the whole, intact. The pelvic and lower limb injuries sustained were therefore a result of the forces involved and interactions with the immediate environment, such as the seat, seat belt and the seat infront. In those areas which had sustained severe damage, and where seating had broken free, interactions with other structures would have taken place but were either unknown or unpredictable or both.

Lower limb injuries following an aircraft accident are common (Mason 1962, 1973, Fryer 1965, Stevens 1970, Horne and Mowbray 1980, Hill 1984). This is the largest series yet documented from one accident. In addition to the direct morbidity and later disabilities, lower limb injuries greatly compromise the survivors ability to escape from a crashed aircraft. Such disabled or trapped victims may survive a crash only to perish from smoke, fire or drowning

(Mason 1973).

4.2 Pelvic and Lower Limb Injuries in Occupants of G-OBME

4.21 Methods

The case records, radiographs and post-mortem reports of all passengers and crew were reviewed. All survivors were interviewed during their hospital stay by either the author or his research assistants (see Appendix 1) and subsequently up to one year following the accident. Initial interviews at three days recorded the incidence and location of soft tissue injuries. Recording these minor surface injuries (bruising, lacerations and abrasions) showed where bodily contact or impact was sustained (Hasbrook 1957), and was therefore an important indicator of injury mechanism. All soft tissue injuries were recorded photographically by the fifth day after the accident.

During follow-up visits, after discharge from the primary treating hospitals, simple anthropometric measurements were made by the author and his colleague G.Tait (see Appendix 1). These are listed in Appendix 3 and include, in particular the buttock-knee length of all those passengers seated in the mid section of the aircraft (Rows 10-20). Measurements were made according to guide lines of Bolton et al (1974). The outcome of the lower leg injuries sustained at nine months were also carried out (Rowles et al. 1991).

All lower limb injuries were categorised into 5 groups:- pelvis; femur; knee; tibia and fibula; ankle and foot. Long bone fractures of the lower limbs were classified according to the 'AO Classification of Fractures' (Muller, Nazarian and Kock, 1987). Talar fractures have been recorded using Hawkins (1970) classification. Open fractures were documented using the Gustilo and Anderson classification (1976). Injuries were correlated with the Injury Severity Scores of the passengers. A break down of all pelvic and lower limb injuries for all occupants and crew is listed in Appendix 2 together with individual Injury Severity Scores. Injury Severity Scores have been calculated using the method of Baker et al (1974) as described in Chapter 3.

For the purposes of the analysis survivors were those occupants that survived to be admitted to a hospital ward (83 patients). The survivors had their injuries well documented with clinical notes, X-rays and other investigations. The non-surviving group - 43 occupants, included four that survived the impact but died soon after removal from the wreckage. Classification of the fracture types in the non-survivors could not be undertaken because of a lack of sufficient X-ray documentation.

The pelvic and lower limb injuries were then related to the seat position and the degree of damage to the aircraft structure and seating. The influence of the crash brace position was also analysed using information from survivors statements, for those seated in rows 10 to 20 where seating had remained attached to the fuselage.

4.22 Results

The Injury Severity Scores (ISS) have been calculated for all 126 occupants of the aircraft. An average ISS of 28 with a range from 1 to 75 was recorded. The average ISS of the 83 survivors of the impact was 15 (range 1-50) and that of non-survivors was 55 (range 21-75).

Table 4.22.1 and table 4.22.2 list lower limb and pelvic injuries for all the occupants and summarises some of the

M1 Aircrash : Lower Limb Injuries
(total of 237 injuries in all occupants)

REGION	NUMBER(%)	PEOPLE(%)	AV.ISS(%)	%COMPOUND
PELVIS	32 (13 %)	32 (25 %)	31	0 %
FEMUR	35 (15 %)	31 (25 %)	35	3 %
KNEE	23 (10 %)	22 (17 %)	24	35 %
TIBIA	69 (29 %)	54 (43 %)	43	64 %
ANKLE	50 (21 %)	42 (33 %)	36	54 %
FOOT	28 (12 %)	23 (18 %)	34	46 %

Table 4.22.1.

Table 4.22.2

Breakdown of Injuries to the Pelvis and Lower Limbs to Occupants of the M1 Aircrash

PELVIC INJURIES:		
Posterior dislocation of hip - with fracture	3	
- without fracture	3	
Acetabular fracture	4(5)*	
Fracture ilium +/- pubic rami	6(7)*	
Pubic rami fracture alone	6	
Diastasis SI jts + diastasis symphysis pubis	3	
Diastasis SI jts alone	2	
Diastasis symphysis pubis +/- fr. pubic rami	5	**32**
FEMORAL:		
Greater trochanter	1	
Intertrochanteric	7	
Diaphyseal	24	
Distal metaphyseal	3	**35**
KNEE INJURIES:		
Femoro-tibial dislocation	5	
Tibial plateau fracture	4	
Posterior cruciate ligament rupture	2	
Lateral ligament rupture	2	
Effusion	3	
Compound laceration	6	
Superior tibiofibular subluxation	1	**23**
TIBIA:		
Proximal	9	
Diaphyseal - single	34	
- comminuted	10	
- segmental	8	
Distal metaphyseal	8	**69**
ANKLE:		
Weber A	5	
Weber B	4	
Weber C	7	
Other intra-articular fractures	4	
Not classified (post-mortem)	24	
Dislocation no fracture	1	
Talar chip fracture	1	
Sprain	4	**50**
FOOT:		
Talar fracture/dislocation - Hawkins type 1	1	
- Hawkins type 2	1	
- Hawkins type 3	6	
Lisfranc fracture/dislocation	11	
Subtalar fracture/dislocation	1	
Calcaneal fracture	1	
Metatarsal fracture	4	
Phalangeal fracture	3	**28**
	TOTAL	**237**

* bilateral injury

data from Appendix 2. A total of 237 pelvic and lower limb injuries were sustained by the occupants (142 by the survivors and 95 in the non survivors). The commonest injuries recorded for the deceased population were to the tibia or shin region (38%) and the ankle region (25%) with many injuries being compound. The least common injuries were those around the knee (5%). For survivors a fairly even spectrum of injuries were seen at each in the lower limb and pelvis, but again the commonest were tibial fractures (32.5%) and the least common injuries around the knee (12.5%).

The injuries to each part of the lower limb are analysed separately in the sections below.

Pelvis

Thirty-two people (25%), 23 survivors and 9 victims, had 32 pelvic injuries. Their average ISS was 31 (survivors 16 - deceased 65). Pelvic fractures occurred throughout the aircraft but most commonly in the centre section (figure 4.22.3) where the seating and restraints remained intact.

In the survivors, there were 8 acetabular fractures, 3 associated with posterior dislocations of the hip. In addition a number of pubic rami fractures and sacroiliac joint injuries recorded. Examination of the seats in which people suffered such injuries revealed indentations in the

Distribution of Pelvic Injuries

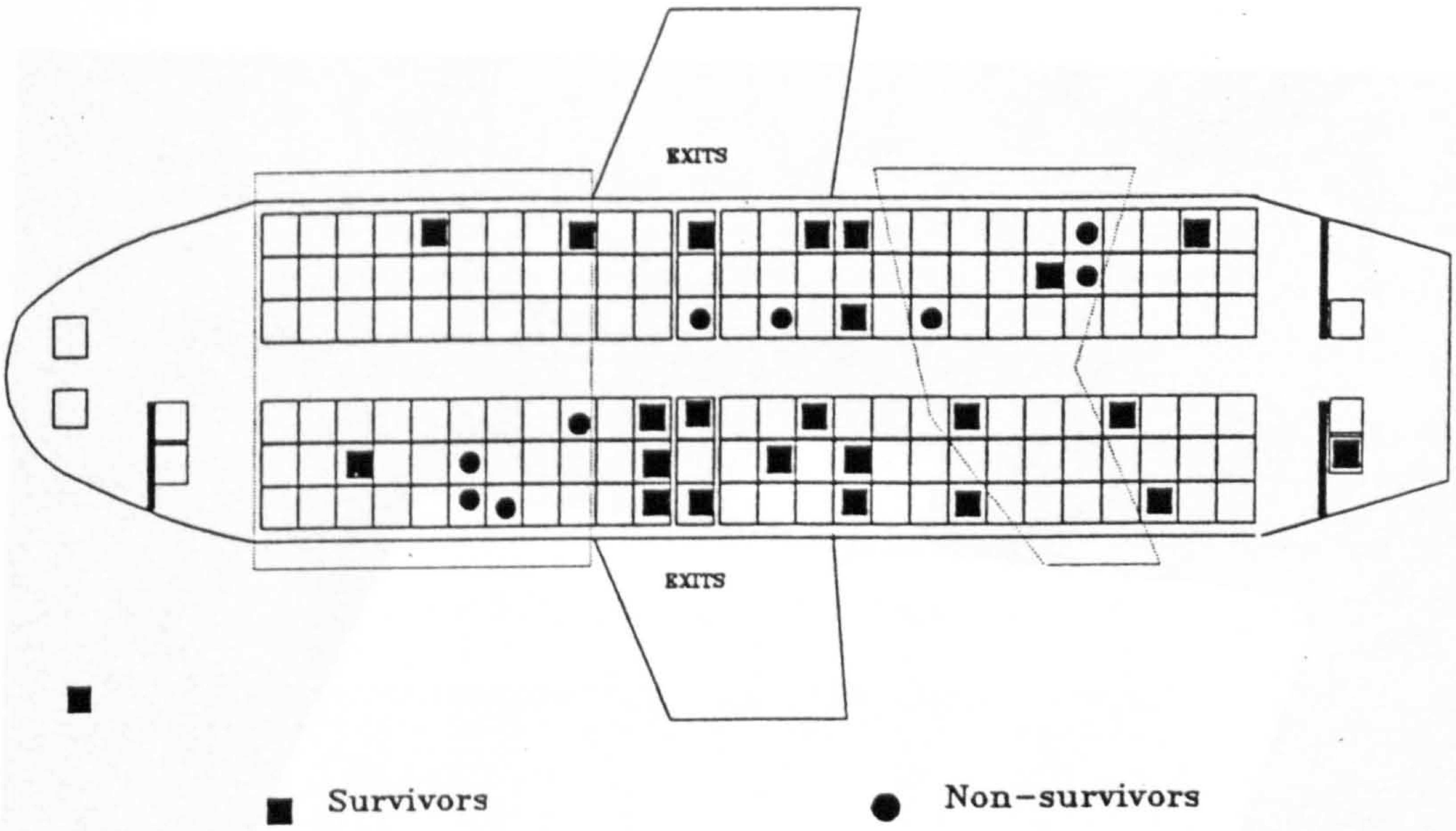


Figure 4.22.3

backs of the seats in front, in the region of the food tray and below (figure 4.22.4). However the knee panel often showed minimal indentation or none at all. The seats also demonstrated some inferior deformation of the seat pan. These indentations were greater degree of severity in the areas of the aircraft which had sustained severe damage with collapse of the seating.

From previous research work on impact biomechanics (reviewed in Chapter 1), different mechanisms of injury may be considered. Pelvic fractures are known to be a result of either direct trauma or as a result of transmission of forces through the femur - the instrument panel syndrome.

Indentations in Seat Back

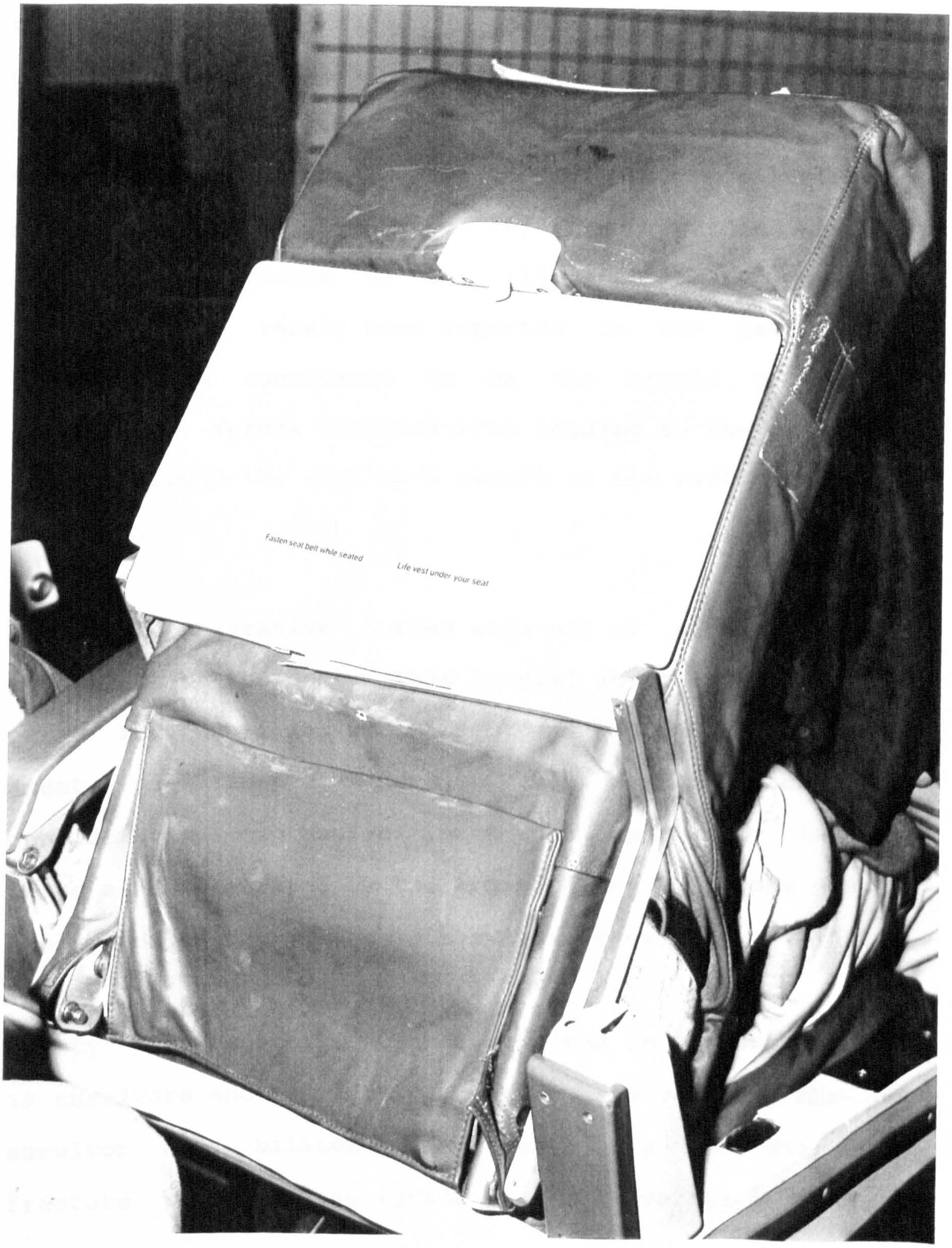


Figure 4.22.4

On impact the passenger is thrown forward, their knees strike the seat in front with the load transmitted down the thigh. This may result in acetabular fractures, hip dislocations and other injuries.

Combinations of sacro-iliac diastasis and pubic fractures or diastases are typical of injuries seen in aircraft accidents. Fractures of the iliac crests seen in this accident have rarely been reported in the past. These injuries are considered to be the result of lateral compressive forces combined with loading to the pelvis and spine through the seat as a result of the vertical forces involved.

Lateral decelerative forces were not of significance in this aircraft accident. Thus lateral impacts as a cause of pubic rami and iliac crest fractures was unlikely to be of great significance. However, it was apparent that considerable compressive loads were imparted by the lap belts as demonstrated by the bruising and abrasions seen in the region of the pelvis in survivors (Rowles et al 1990).

Femur

Thirty five femoral fractures occurred in 31 (25%) people, 19 survivors and 12 victims. Three survivors and one non survivor had bilateral fractures. Only one diaphyseal fracture was compound (grade 2). The average ISS was 35 (survivors 20 - deceased 59).

The distribution of femoral fractures is seen figure 4.22.5. This seat plan demonstrates that femoral fractures occurred throughout the fuselage with some sparing of the last few rows in the aircraft.

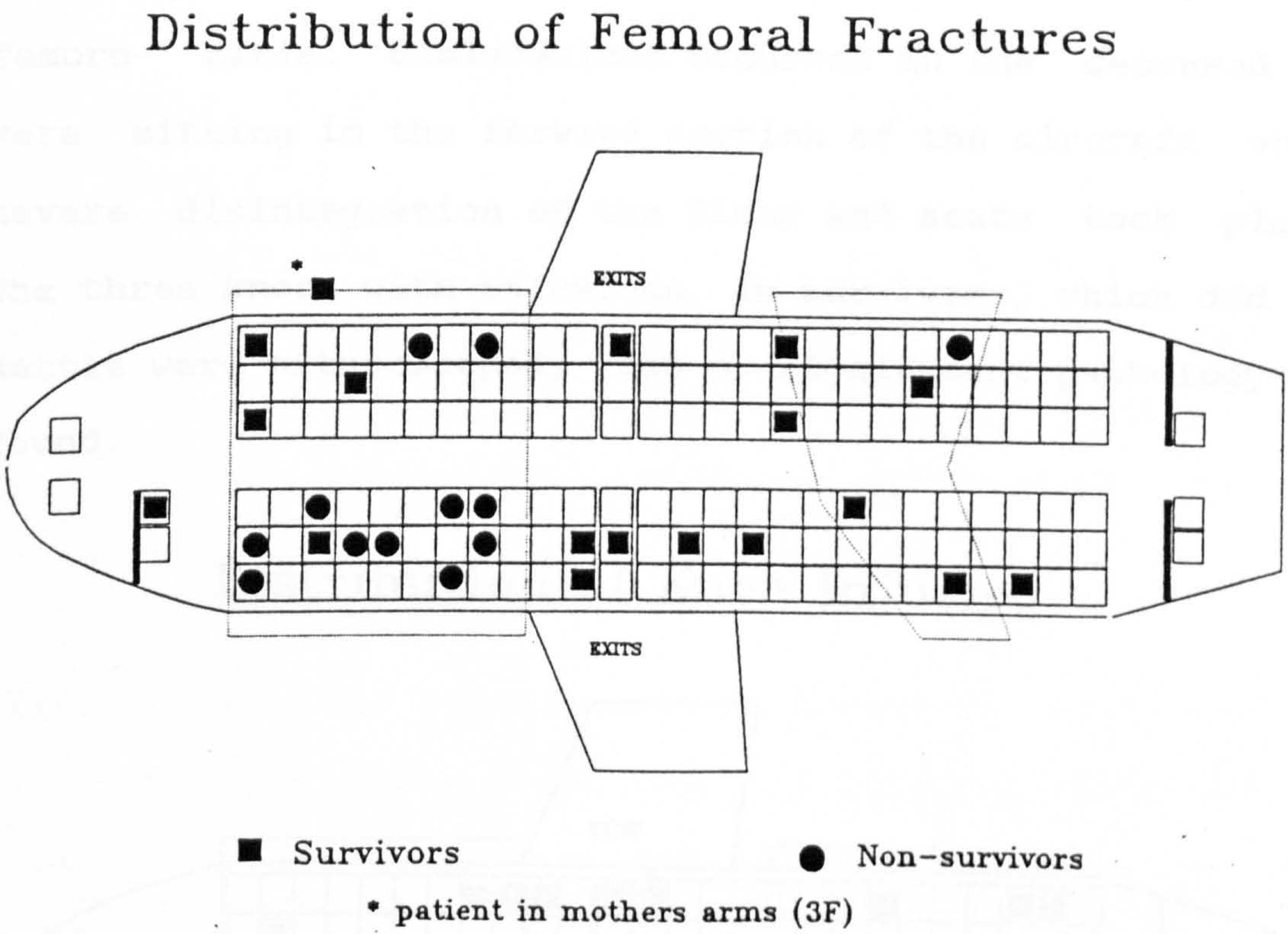


Figure 4.22.5

The mechanism of femoral injury due to impact, defined as a result of automotive research, has found general acceptance in the aviation industry. This is the well described instrument panel syndrome, were primary axial loading of the femur results in a bending failure of the bone in the thigh. This mechanism was thought to be supported by the witness indentations seen in the rear of the seats, and which were especially common in those areas of the aircraft that sustained severe damage.

Knee

Twenty two (17%) people, 17 survivors and 5 victims suffered 23 knee injuries. The type and distribution of the injuries are illustrated in figure 4.22.6. All five femoro- tibial dislocations occurred in the deceased who were sitting in the forward section of the aircraft where severe disintegration of the floor and seats took place. The three knees with effusions, in survivors, which did not settle were arthroscoped, but no significant pathology was found.

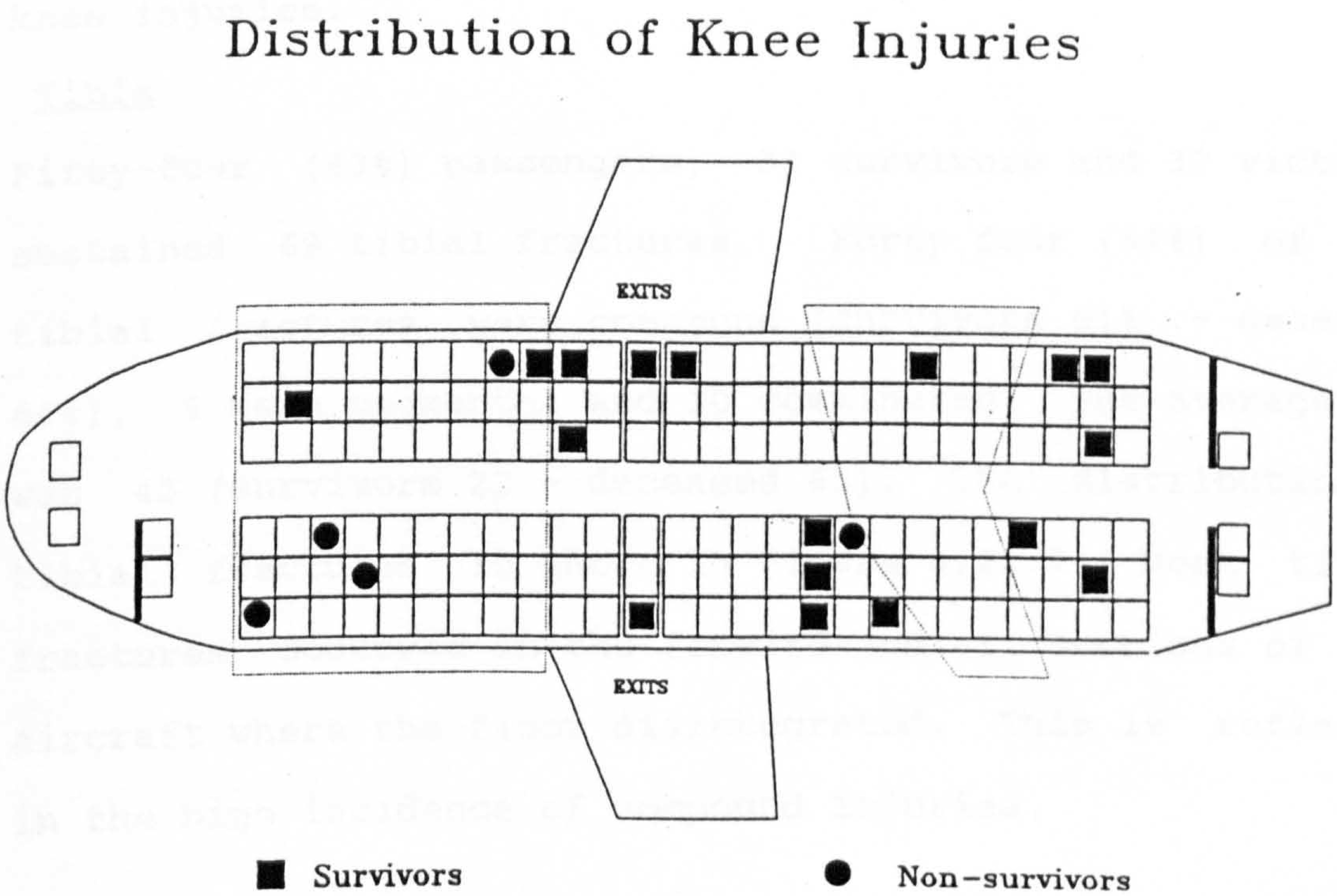


Figure 4.22.6

Viano et al (1978) demonstrated that bolster impacts to the knee, occurred as a result of impact with the instrument panel in automobiles, resulted in a variety of injuries around the knee depending on where the joint was struck. Implicit to these injuries was impact of the knee against a bolster. A number of the knee injuries seen in the occupants of the aircraft could be explained by such a mechanism. Examination of the lower part of the backrest of the seats of G-OBME demonstrated indentations and when the back rest was tilted forward for inspection poorly protected bolts and edges level with the knee of the occupant behind were identified [See pictures]. These edges and projections may have accounted for some of the open knee injuries.

Tibia

Fifty-four (43%) passengers, 31 survivors and 38 victims, sustained 69 tibial fractures. Forty four (64%) of the tibial fractures were compound (survivors 61% - deceased 66%), 8 were segmental and 10 comminuted. The average ISS was 43 (survivors 22 - deceased 63). The distribution of tibial fractures is shown in figure 4.22.7. Most tibial fractures occurred in the forward and aft sections of the aircraft where the floor disintegrated. This is reflected in the high incidence of compound injuries.

In those areas of the aircraft that remained structurally intact the injuries seen were concluded to have resulted

Distribution of Tibial Fractures

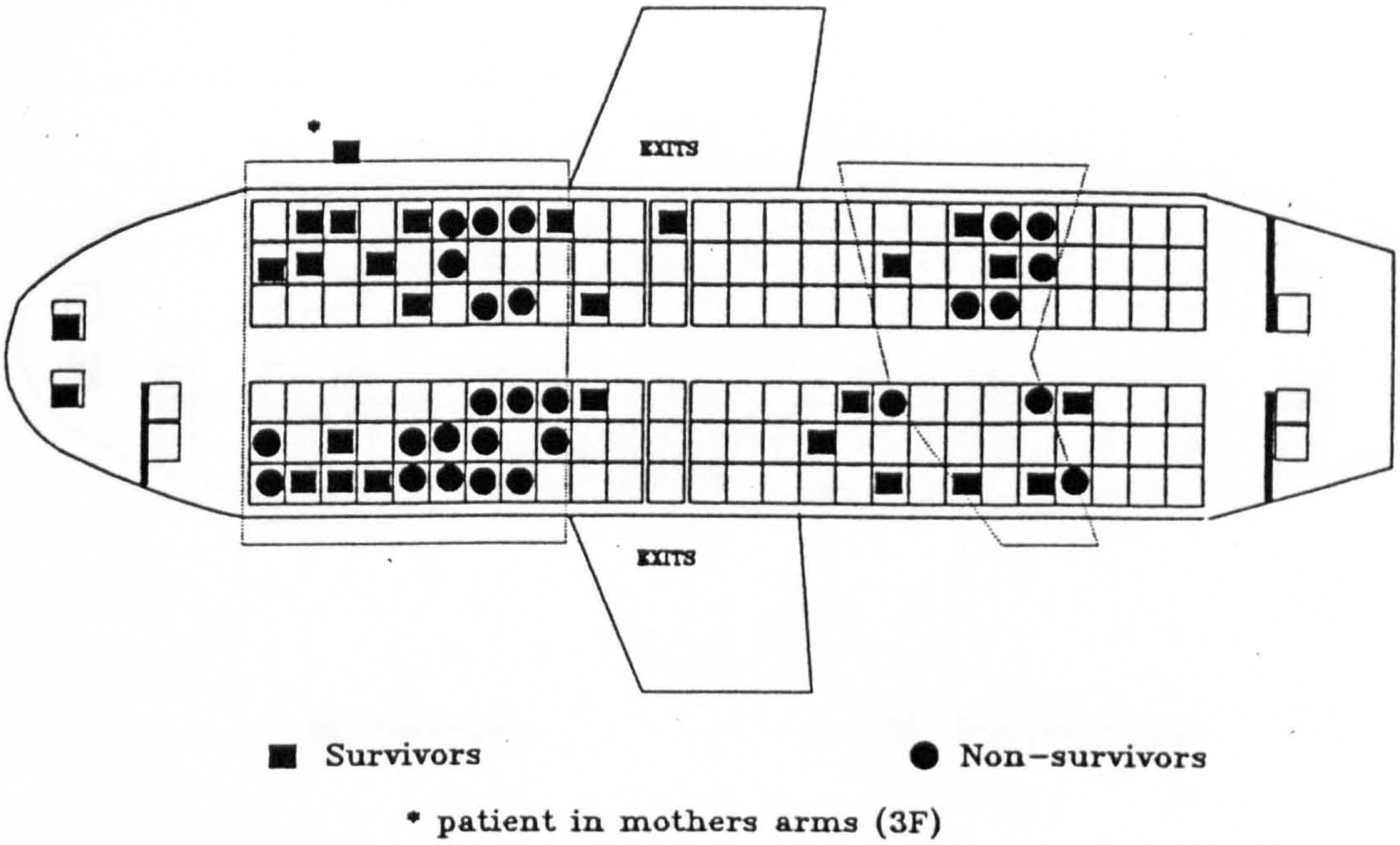


Figure 4.22.7

from flailing of the lower limbs as described by Swearingen et al (1962).

Ankle and foot

Forty-two (33%) people, 26 survivors and 24 victims, suffered ankle injuries, 8 of them bilateral and 27 (54%) compound. The average ISS score was 36 (survivors 20 - deceased 55). Twenty-three (18%) occupants, 22 survivors and 6 victims, sustained 28 foot injuries, 13 (46%) of which were compound. The average ISS was 34 (survivors 20 - deceased 75).

Talar fracture-dislocations (Aviator's Astragalus) and

Distribution of Ankle Fractures

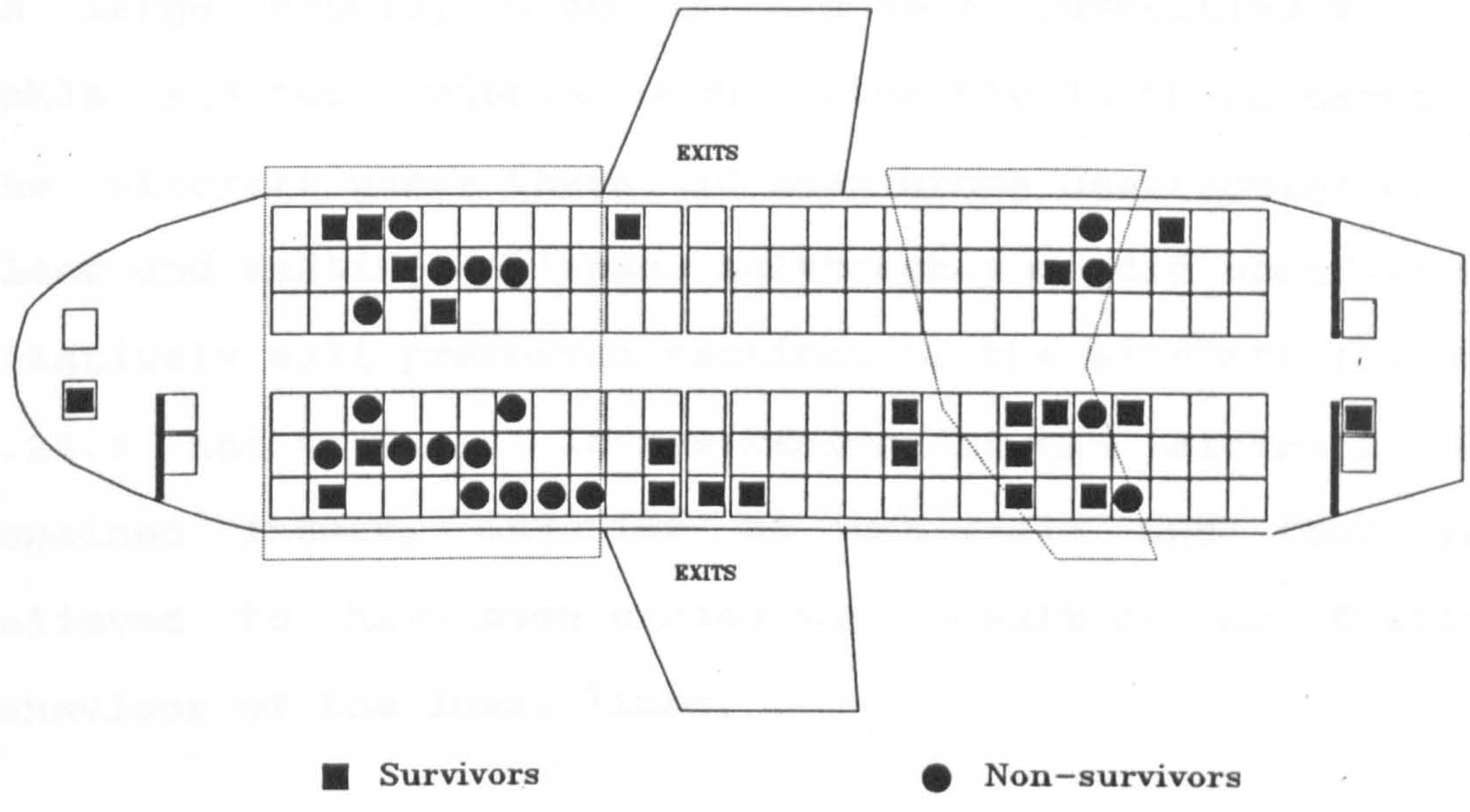


Figure 4.22.8

Distribution of Foot Injuries

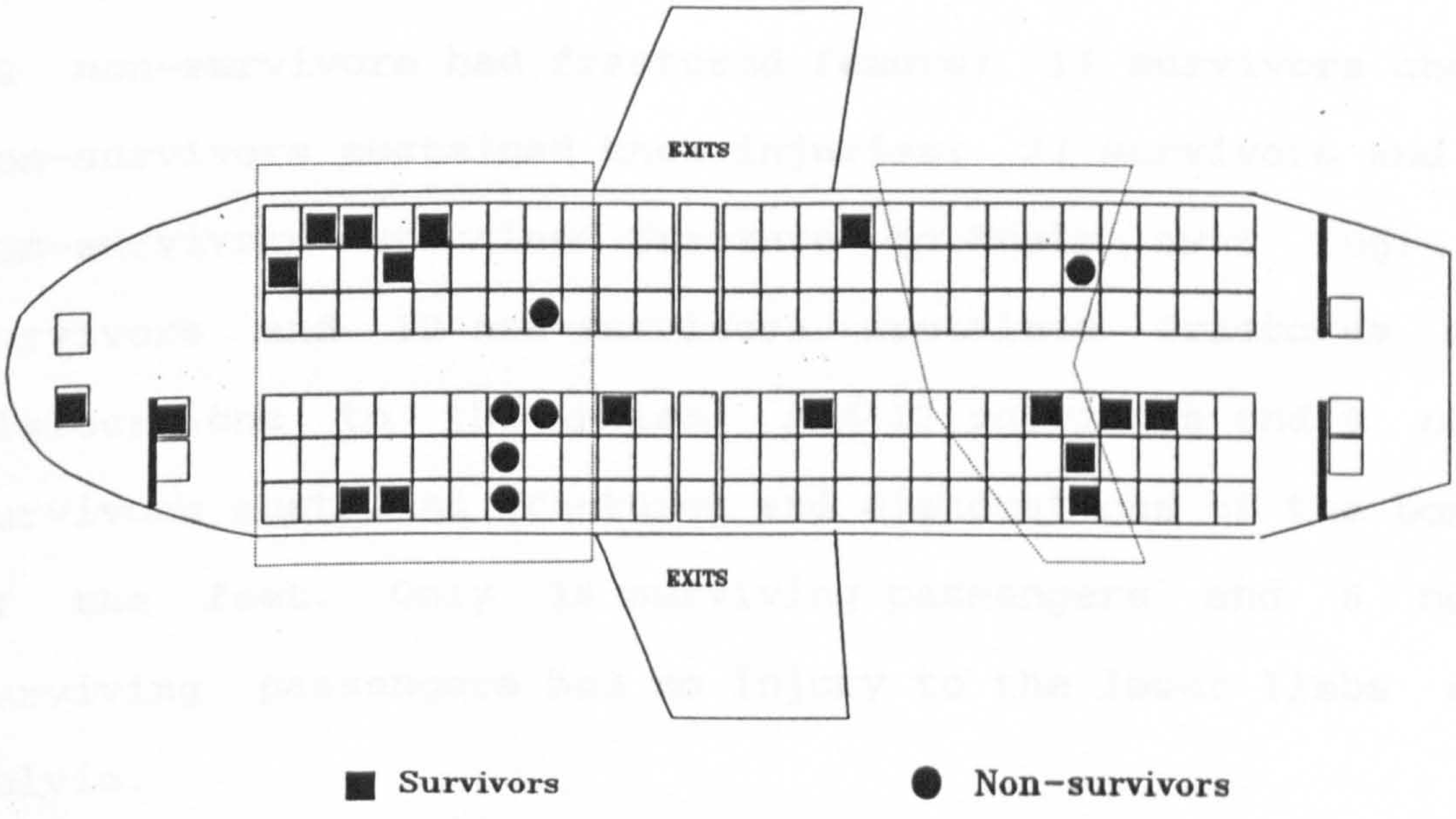


Figure 4.22.9

Tarso-metatarsal (Lisfranc) fracture-dislocations were seen in large numbers, 8 and 11 were seen respectively. The ankle and foot injuries occurred mainly in those parts of the aircraft where there had been gross destruction of the floor and seating collapse, although some did occur in the relatively well preserved sections of the aircraft (figures 4.22.8 and 4.22.9). In the regions of the aircraft that remained intact, injuries to the ankle and foot were believed to have been caused as a result of the flailing behaviour of the lower limbs.

4.23 Discussion

As a result of the forces in the crash impact a wide variety of injuries were sustained to the pelvis and lower limbs of the occupants. Twenty three survivors and 9 non-survivors sustained pelvic injuries; nineteen survivors and 12 non-survivors had fractured femurs; 17 survivors and 5 non-survivors sustained knee injuries; 27 survivors and 27 non-survivors sustained fractures to their lower leg; 23 survivors and 19 non-survivors sustained fractures and dislocations to the ankles; and 17 survivors and 6 non-survivors sustained fractures and dislocations of the bones of the feet. Only 18 surviving passengers and 6 non-surviving passengers had no injury to the lower limbs and pelvis.

From the injury severity scores it can be seen that the deceased group sustained severe multisystem injuries in addition to limb injuries. The regions of high mortality have resulted primarily from the collapse of the airframe.

Mechanisms of injury of the pelvis and lower limb have been deduced from an analysis of the injuries of the survivors and the damage to the seats and aircraft structure as well as information provided by an extensive literature review. The floor distortion and seat disruption had an obvious effect on the incidence of injuries to the lower limbs of the occupants. This is especially true of injuries distal to the knee with a particularly high incidence of open (compound) injuries. This indicates a severe crushing type of injury experienced by the occupants in those regions of the aircraft.

The following mechanisms for the pelvic and the lower limb injuries can thus be suggested:- Iliac wing fractures were caused by deceleration of the pelvis against the lap belt producing compression of the iliac blades. Fractures of the pubic rami and diastasis of the sacroiliac joint, result from a combination of deformation of the seats, compressive forces produced by lap belts and transmitted forces. In the cases of crew members seated in rear facing seats fracture of the pubic rami were a result of a direct blow.

The knee-femur-pelvis (instrument panel syndrome) mechanism of injury could explain the mechanism of many of the lower limb injuries sustained by the occupants. In particular the acetabular and hip dislocations, the femoral fractures and the injuries around the knee. On impact a passenger, even with a tight lap-belt, is propelled forwards, the knees strike the lower part of the seat in front resulting in injury to the tibia, knee and distal femur. Forces are transmitted along the femur, driving it backward into the pelvis causing femoral and acetabular fractures, hip dislocations and pelvis shear fractures.

Flailing of the lower limbs under the seat in front, as described by Swearingen et al (1962), also explains the likely mechanism for shin, ankle and foot injuries. However the mechanisms described above only hold true if the seating and restraint systems remain intact.

Femoral fractures were distributed throughout the aircraft but the mechanism of injury appeared to vary with seat position. Passengers seated in the forward and aft sections which were the most badly damaged areas, suffered bending and comminuted type fractures as a result of crushing and interactions with seating. A segmental fracture of a femur might have resulted from the limb sliding under the seat ahead with bending of the thigh over the seat edge and under the seat in front. Forward translation of the femur on a tibia wedged behind seats would result in total disruption of the knee.

The incidence of tibial, ankle and foot injuries is known to be influenced by collapse of the seating and destruction of the floor of the aircraft. The majority of lower limb injuries occurred in those regions which had sustained severe damage. The basic mechanisms of injury described being modified by the destruction of the floor, with resulting trapping and crushing of the lower limbs. Many of these injuries were open or compound injuries with bones sticking through the skin.

The mechanisms of some tibial, ankle and foot injuries can be inferred from the fracture types. For example, posterior malleolar fracture dislocations, were caused as a result of the tibia being driven forward on a foot planted on the floor; tibial plateau fractures were caused by an eccentric axial load on a planted straight lower limb. Eight talar fracture-dislocations, the classic "Aviator's Astragalus", and 11 Lisfranc tarsometatarsal fracture-dislocations were caused as occupants were thrown forward in their seats, catching their feet in the wreckage of the aircraft. This mechanism was described in 1980 by Horne and Mowbray as being the cause of the foot and ankle injuries seen in the accident of a DC-9 in 1978.

This accident has demonstrated that sitting in an area of the aircraft which remained intact confers survival advantages but does not necessarily infer that you escape without injury. The incidence of pelvic injuries was particularly high in the mid section of the aircraft, whereas, tibial, foot and ankle injuries were more common in those areas that sustained severe structural damage. Pelvic injuries may have been more frequent in the centre section because of under reporting of injuries in deceased occupants as previously discussed or because the forces transmitted to the occupants in the mid section were not deminished by failure and collapse of seating and fuselage.

It is of special interest to compare the injuries suffered by the cabin crew who were seated in rear-facing seats with upper body restraints. Several of the cabin staff suffered lower limb injuries despite facing aft and thus the proposed mechanisms would not be expected to hold true. The causes of their injuries, revealed by examination of the aircraft and statements from the staff, were the consequence of the collapse of cabin fittments and being struck by cabin debris.

It is now well recognised that effective body restraint will minimise injury and prevent death. Mosely and Zeller (1958) found that "dislodgement of the seat and passenger was the most prevalent injury factor" in passenger aircraft crashes. This is born out in the Kegworth accident, however even those areas where seating and restraint remained intact, significant injuries to the lower limbs were still prevalent. Had there been an outbreak of fire many passengers would have been unable to escape from the aircraft because of their limb injuries.

4.3 Biomechanical Features of Pelvic and Limb Injuries in Survivors Seated in the Mid Section (Row 10 - 20) of G-OBME

It is postulated that the majority of lower limb injuries in the M1 aircraft accident were caused primarily by the impact of a passenger into the back of the seat in front, or as a consequence of the severe structural damage sustained by the aircraft. A wide spectrum of injuries are demonstrated by similar postulated impacts throughout the aircraft.

The currently accepted mechanism of injury, well described in the automobile industry is described below. On impact a passenger seated in an intact region of the aircraft is propelled forward and the knees strike the bottom of the seat ahead, causing injuries to the knee, upper tibia and lower femur. The impact forces are transmitted up the femur, driving it backwards into the pelvis. This leads to femoral shaft fractures, hip dislocations, acetabular fractures and pelvic shear fractures.

For this scenario to be applicable to this aircraft accident evidence of knee contact with the seat in front is required for those passengers demonstrating injuries sustained by this mechanism. Distal tibial fractures, and fractures and dislocations of the foot and ankle, were accepted as resulting from the disintegration of the cabin floor in the forward and aft sections of the aircraft overlying the cargo hold and a consequence of flailing of the lower limbs.

This section reviews pelvic and lower limb injuries in those occupants who survived the aircrash seated in the central section of the aircraft. The position adopted at the time of impact has been reviewed, simple anthropometric measurements have been taken and soft tissue injuries documented. For this group of individuals a good data set is available with X-ray documentation of all their injuries.

4.31 Methods and results

Table 4.31.1 reports the pelvic and lower limb injuries in the occupants seated in rows 10 to 20, a region of the aircraft where the seats remained attached to the floor and where a survivable occupant environment was maintained. It can be seen that a wide variety of injuries were sustained. The femoral fractures seen in this region were more frequently documented as proximal femoral fractures.

Table 4.31.1

Pelvic and Lower LImb Injuries in Survivors Seated in Rows 10 - 20

SEAT NUMBER	BRACE POSITION	SOFT TISSUE WITNESS	FRACTURES AND AO CLASSIFICATION	HEIGHT cm	WEIGHT kg	BUTTOCK KNEE cm
10C	Unknown	Rt knee Lt foot	Rt os.calsis + open MT's Lt 1st MTPJ	168.5	67	61
10D	Partial brace	Rt shin Rt thigh Lt knee	Rt tibial plateau - 41B1 Lt Tibia/fibula - 42B2	185	66.7	59
10F	Partial brace	Rt shin Lt knee Feet	Rt posterior cruciate	159	49.2	55
11A	Braced Legs forward	None	Rt femur shaft - 32A2	165	95.5	-
11B	Braced Legs upright	None	Rt femur shaft - 32A2 Lt femur shaft - 32B2 Lt medial malleolus-43B2 Rt SPR + diastasis SIJ	148	58	59
11C	Unbraced Legs upright	None	Rt iliac creast Rt SPR + IPR	166	54	59

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11D	Partial brace Legs upright	None	None		169	73	57
11F	Unknown	Effusion Rt Knee Rt + Lt shins	None		188	101	-
12A	Unknown	Rt knee Lt knee	Rt ankle Rt iliac crest	- 44C2	157	55.5	57
12B	Braced Feet under own seat	Rt + Lt feet	Lt intertroc.femur	- 31A2	164	63	62
12C	Unknown	None	Rt iliac crest		159	55	55
12F	Seat broke free Braced	Lt upper shin Rt knee	Rt acetabulum Lt acetabulum Rt femur Lt tibia	 - 32B2 - 42A2	182	95	58
14A	Braced Lt leg back Rt leg forward	Lt shin Rt ankle	None		170	76.5	63
14C	Unknown	Lt + Rt heels	None		168	59.5	57
14D	Unknown	None	None		178	55	-
14F	Seat broke free Braced Legs upright	Lt shin Effusion Lt Knee	None		184	84	64
15A	Unknown	Rt + Lt shins Rt knee	None		170	68	-

15B	Unknown	None	Lt intertoc. femur- 31A2 Rt SPR + IFR	165	52	-
15C	Braced Legs upright	None	None	167	69	56
15F	Braced Legs forward	Rt knee	None	169	83	56
16A	Unknown	Rt knee Lt knee	None	162	76.5	57
16C	Braced Feet under own seat	Lt knee	Rt metatarsals	162.5	57	54
16D	Unknown	Rt knee Lt shin	None	161	61.4	55
16F	Unbraced Legs upright	None	Lt acetabulum	166	60.5	58
17A	Braced Legs forward	Rt shin Lt shin	Rt iliac crest	183	78	59
17B	Unknown	Lt upper tibia	Rt acetabulum Lt femur + Disln - 31A2 Rt tibia/fibula - 42C3 Lt tibia - 41C2	173	63.5	-
17D	Braced Legs back	None	Rt + Lt iliac craest	182	76	61
17F	Unbraced	None	Rt SPR + IPR Rt talus Rt Lisfranc	160	54	-

18A	Unknown	Rt knee Lt shin	Rt tibial plateau	- 41B1	148	65.5	52
18B	Partial brace Legs upright	Effus'n Lt knee Lt shin	Lt ankle Rt ankle	- 43B2 - 44C1	169.5	64.6	56
18C	Partial brace Legs upright	Rt shin Effusion Lt knee Sprain Rt ankle	L tibia	- 42B2	170	82.5	57.5
18D	Braced Legs upright	Rt knee Lt knee + shin Rt foot	Lt femur	- 32B1	178	63.5	56
18F	Partial brace	Rt knee Lt + Rt shin	Lt femur	- 32C2	-	-	-
19A	Partial brace Legs forward	Rt shin Lt heel	Rt tibia	- 42A3	180	67	57.6
19E	Braced Legs forward	Rt knee	Rt tibia	- 42B2	178	63.6	56
19F	Braced Legs upright	Rt knee Lt + Rt ankles	None		181	75	62
20A	Partial brace	Rt knee	Posterior disl'n Rt hip		176	89	-
20C	Braced Legs upright	Lt shin	Rt femur	- 32A3	180	82.7	56

The seating and restraint systems in the mid section of the aircraft remained on the whole intact. In rows 12 and 14 the pitch of the seats was increased from 32" to 38". Two seats 12F and 14F broke free from the remaining seats in the row, and the occupants were ejected from the aircraft. Figure 4.31.2 illustrates the basic design of seat rows. The rows of seats are supported by legs that are eccentrically placed such that the seats adjacent to the fuselage have a greater 'overhang' or unsupported length.

Passenger Triple Seat

(From AAIB 1990)

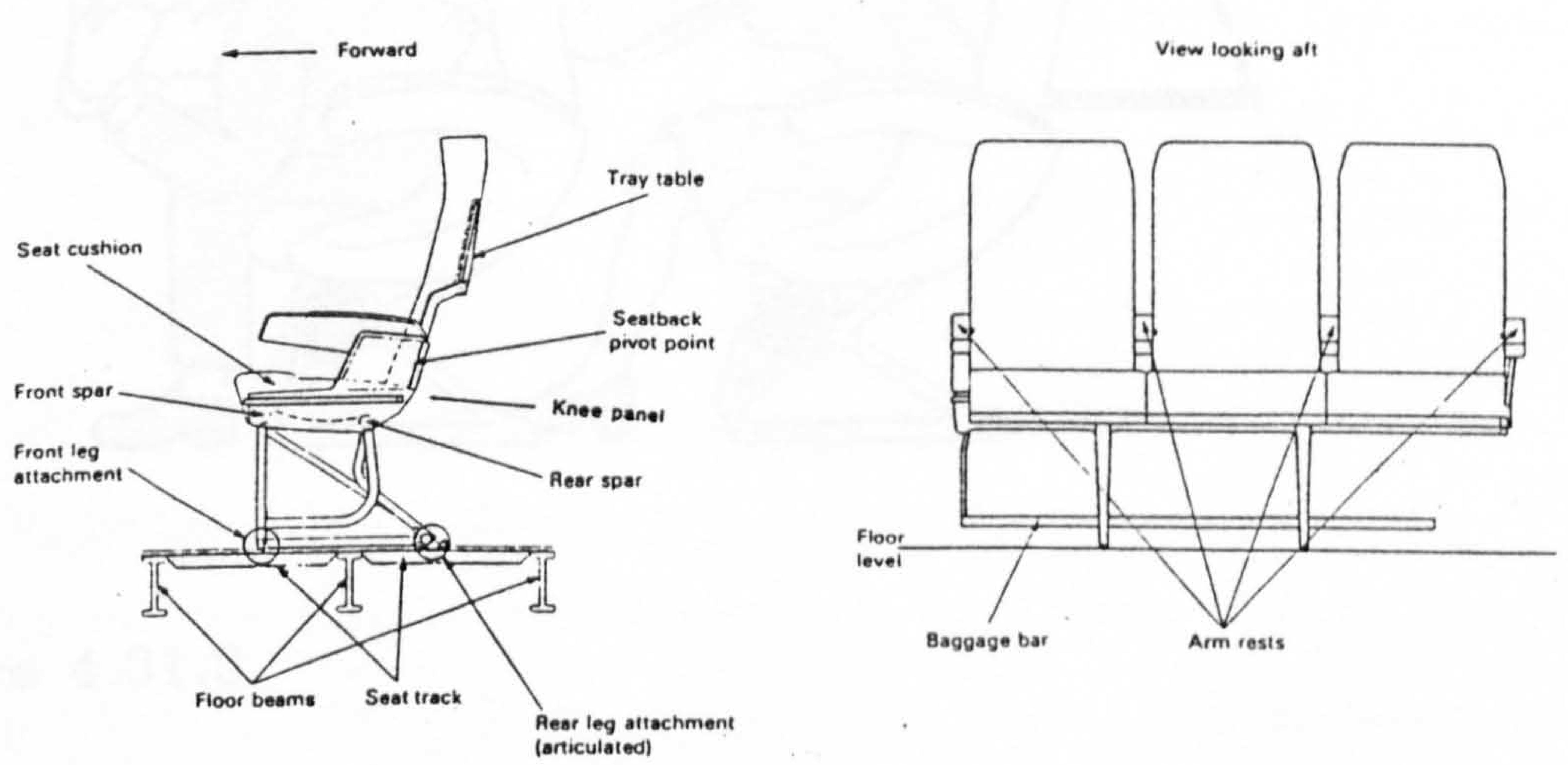


Figure 4.31.2

Brace position

Passengers seated in the mid section of the aircraft were asked to recall the position they adopted at the time of impact, in particular the placement of their lower limbs.

If the passengers assumed a position as recommended by the British Midland Safety Instruction Card (figure 4.31.3) they were classified as adopting a 'braced' position. Those passengers failing to adopt such a position but bracing in some other way were classified as 'partially' braced. Those passengers who remained seated upright were recorded as

Brace Position

From British Midland Saftey Instruction card 1989

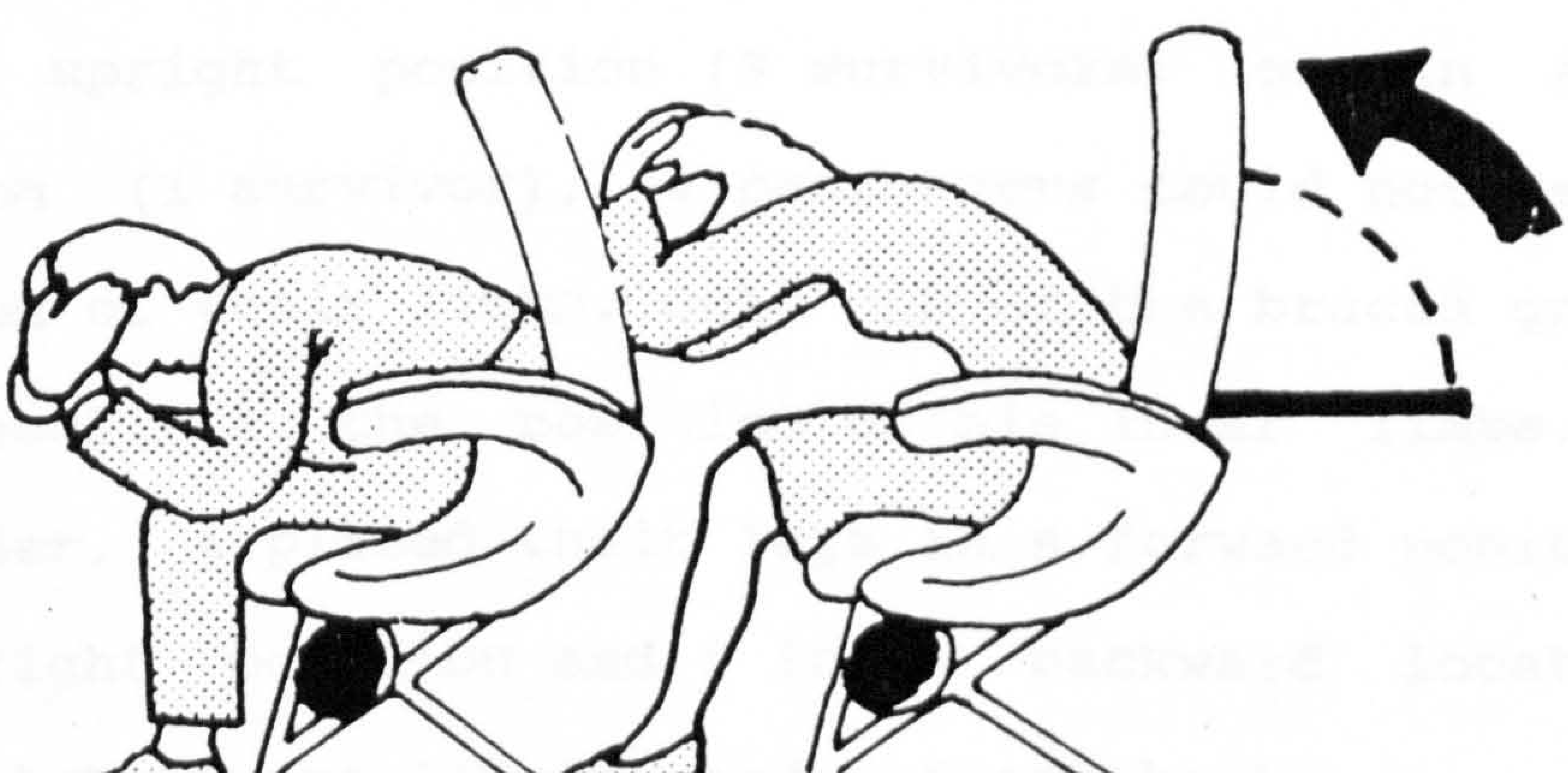


Figure 4.31.3

Three positions the lower limbs could adopt were identified:- legs forward, the legs resting on the floor in front of a vertical axis drawn through the knee joint; legs upright, the feet resting on the floor with the knees approximately vertical; and feet under the occupants own seat. The position of the legs was not recorded if the occupant could not accurately recall their foot position.

Thirty eight people who survived and were admitted to a hospital ward were reviewed. Twelve occupants had no recollection of the accident and were therefore unable to give an account of the crash position adopted. Of the remainder, 3 passengers were unbraced, 8 were partially braced and 15 were braced. Of the occupants who were unbraced 2 placed their legs in an upright position; partial braced individuals either placed their lower limbs in an upright position (3 survivors) or in a forward position (1 survivor), 4 passengers could not recall the position of their limbs. Only one of the braced group could not recollect the position of his lower limbs. Of the remainder, 4 placed their legs in a forward position, 6 in an upright position and 3 in a backward location. One occupant held one leg forward and one back.

Brace position and injuries

Because of the small sample numbers and the difficulties in the patient's recollections of the positioning of their lower limbs at the time of the impact, no statistically significant associations of the crash position of the lower limbs with injuries to the lower limb and pelvis emerge, however the range of positions in which the lower limbs were placed is demonstrated.

Six of the nine femoral fractures (66%) occurred in individuals who adopted a brace position, with 1 occurring in a partial braced occupant and 2 in the unknown group. Comparing the braced population with the rest demonstrates no significant increase in fractures in those that were known to be braced (Fisher exact probability test p=0.146).

Soft tissue witness mark's and fractures

Fractures were recorded as previously described. Abrasions, lacerations and bruising to the lower limbs, were recorded and were categorised as soft tissue witness mark or evidence of impact (Hasbrook 1957).

Twenty six patients demonstrate soft tissue injuries to their lower limbs and 26 sustained a pelvic or lower limb fracture. Of those passengers who sustained fractures, pelvic injuries were the commonest, with a high incidence of femoral fractures. Of those patients who sustained femoral fractures four occupants out of six were seated in the central seat within a row. This has been compared with five femoral fractures in the remaining thirty two patients not seated in a central seat in a row. The increased incidence of femoral fracture in the centre seat of a row was found to be significant (Fisher exact probability test, p = 0.0398).

Soft tissue injuries were present on the shins and feet, reflecting contact of these areas at impact and therefore flailing of the lower limb under the seat in front. This was also indicated by the tibial fractures, ankle and foot injuries. Impacts to the ipsilateral shin occurred in association with injuries to the knee joint (effusions of joints or rupture of the posterior cruciate ligament) as has been described in bolster impacts to the knee and tibia (Viano et al 1978).

Soft tissue injuries were also sought and some were identified around the knee and were considered to be an important witness of impact of the knee and possible axial loading of the femur. Analysis of the femoral fracture types seen demonstrated <u>spiral fractures</u> - indicating torsion, bending fractures with butterfly fragments, <u>oblique fractures</u> - suggesting axial loading, and <u>segmental fractures</u>. The wide variety of fracture types seen suggest no one specific mechanism was applicable to all individuals.

The occupant in seat 12F deserves mention, this seat broke free. The fracture of his right femur was a bending wedge type fracture with a central hip fracture dislocation above. On the left side he had a transverse fracture of the acetabulum. He also had major lacerations of his right knee and left upper shin. This occupant appears to have been thrown forward into the seat in front impacting his knees and upper shin on the seat infront. The load was then transmitted through his femurs and resulted in the injuries noted above.

Table 4.31.4 illustrates the presence of soft tissue witness around the knee in relation to the presence or

Soft tissue witness and the presence of a fracture associated with axial loading

Fracture associated with axial loading	Presence of soft tissue witness around the knee: YES	NO
YES	4*	7
NO	12	15

Fisher p=0.6849

* 1 soft tissue witness on contralateral limb

Table 4.31.4

absence of injuries that may result from axial loading of the femur (femoral fractures, hip dislocations and acetabular fractures). Of those 11 patients whose fractures might have been attributable to axial loading of the femur (the knee-femur-pelvis complex) only three patients demonstrated a soft tissue lesion of the ipsilateral knee (one patient with a soft tissue injury on the contralateral knee). Twelve patients demonstrated a soft tissue witness mark but no fracture and fifteen patients demonstrate no soft tissue injury or fracture. Thus 16 (42%) of the occupants sustained soft tissue injuries around the knee and 22 (58%) did not. There is no apparent relationship between the incidence of injuries associated with axial loading of the femur and the presence of a soft tissue witness mark to the knee (Fisher exact probability test, p = 0.6849)

Anthropometry

Measurements made on patients are recorded in Appendix 3. Of particular interest is the 'buttock to knee' length. This is the distance recorded in the seated position from the anterior surface of the patella to the back of the buttock. Table Appendix 3.3 demonstrates the average anthropometric measurements made on individuals seated in the centre section of the aircraft. Measurements were not obtained in nine of the 38 occupants seated in the mid section of the aircraft.

The average occupant standing height was 170 cm (160 cm in females and 175 cm in males) and their weight was 69 kg (58 kg in females and 74.5 kg in males). For the buttock knee lengths the average length was 57.8 cm (56.4 cm females and 58.5 males). These measurements compared favourably with the design specification of the hybrid III anthropomorphic test device (Foster et al 1977, personnel communication D

Anton 1990).

Anthropometric measurements and injuries

The knee-femur-pelvis (instrument panel) mechanism requires that the knee must impact with the seat in front. Because of individual variation one would expect those individuals with longer femurs (therefore a greater buttock-knee distance) to demonstrate an increased incidence of soft tissue injury around the knee. Table 4.31.5 and 4.31.6 illustrates the occurrence of soft tissue injuries around the knees of occupants, or the presence of a femoral fracture in relation to the average buttock thigh length (57.8 cm).

Table 4.31.5 demonstrates that buttock knee length is apparently unrelated to soft tissue injury with four out of twelve patients, with a 'greater' than average buttock knee length, demonstrating a witness mark. In the 'less' than average group nine out of seventeen occupants demonstrate a witness mark (Chi-square p = 0.505).

For fractures associated with axial loading of the femur (Table 4.31.6), 4 (33%) of occupants with above average buttock knee length sustained a fracture, whereas in the less than average buttock knee length only 2 (12%) sustained such an injury. However this difference is not statistically significant (Fisher exact probability test p = 0.344).

Buttock - knee lenghth and lower limb injuries (soft tissue injuries)

		Buttock - knee lenghth average 57.8 cm		
		GREATER	LESS	UNKNOWN
Soft tissue injury around knee	YES	4	9	3
	NO	8	8	6

Chi-square p=0.505

Table 4.31.5

Buttock - knee lenghth and lower limb injuries (fracture associated with axial loading)

		Buttock - knee lenghth average 57.8 cm		
		GREATER	LESS	UNKNOWN
Fracture associated with axial loading	YES	4	2	5
	NO	8	15	4

Fisher p=0.344

Table 4.31.6

4.32 Discussion

Injuries sustained by the pelvis and lower limbs of the 38 occupants seated in rows 10 - 20 in the mid section of the aircraft have been reviewed. This region of the aircraft remained on the whole intact with the floor seating and restraint systems retaining there integrity. Injuries sustained were therefore a result of both the transmission of the forces involved in the impact and the interactions with the immediate surroundings.

In the previous section, review of lower limb and pelvic injuries sustained by all occupants of G-OBME, mechanisms of pelvis and lower limb fractures have been suggested. Fracture types indicate that the forces producing fractures of the lower limb include axial loading, torsional and bending forces.

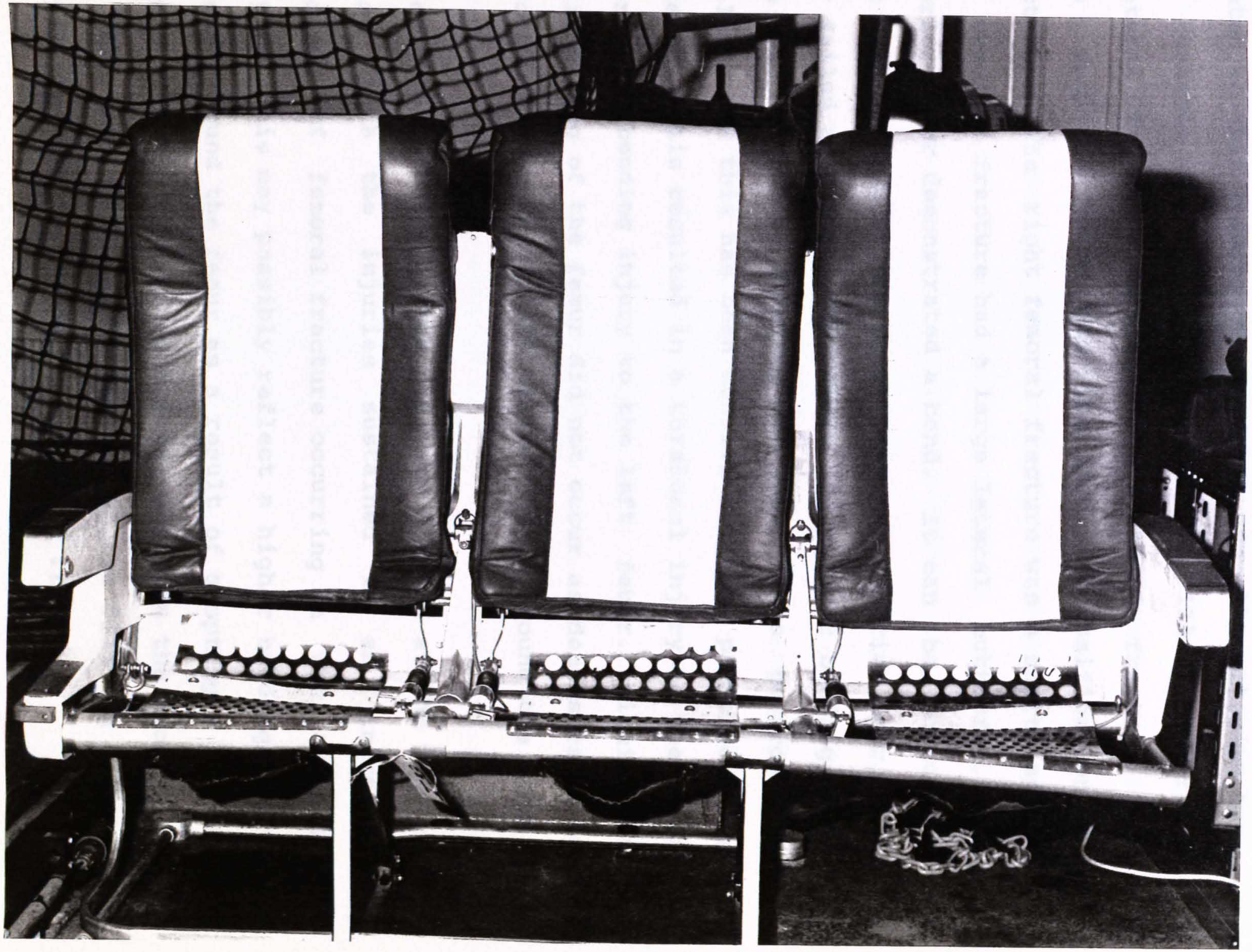
Flailing of the lower limbs has been demonstrated as occurring following aircraft accidents and results in many of the injuries to the tibia, ankle and foot seen in the victims of aircraft accidents. Examination of the fracture types in the lower leg, has demonstrated that some of the injuries have resulted from eccentric axial loads, with or without rotation (some ankle fractures and tibial plateau fractures). For axial loading of the tibia to occur the foot must be planted on a flat surface analogous to the floor pan of an automobile. In the presence of lower limb flailing axial loading of the shin cannot occur. This suggests that flailing of the lower limbs did not occur in all occupants.

Axial loading of the femur has been suggested as being important in the causation of some femoral and hip injuries. If this is the case then witness marks of impact should be apparent around the knee joint. This situation has not been proved for the victims of the M1 aircrash seated in the mid section of the aircraft. Further analysis has demonstrated that the presence of a soft tissue injury to the knee was not associated with an individual's buttock-knee length. However buttock knee length may be related to the incidence of fractures to the femur and hip in the absence of soft tissue witness marks, although numbers are to small to reach statistical significance.

From examination of the back of the seats in front, some indentation of the knee panel is evident but was only of a minor degree. This may represent contact with the seat in front by the occupant seated behind. However it is not necessarily true that the indentation was caused by impact of the knees but could have been some other part of the anatomy. Gadd et al (1970) performed drop tests on soft tissues of the scalp and fatty tissue of the upper arm and demonstrated 120 lbs force (533 N) was required to cause a visible crush injury, by compressive parallel impact in cadavers. This is a degree of magnitude 10 times below the injury threshold for axial loading injuries. It therefore seems likely that axial loading of the femur (the knee-femur-pelvis mechanism) as described in the automobile industry may not apply to all the occupants seated in the mid section G-OBME.

Review of the damage to seating has revealed that the anterior spar (anterior cross member) of the seat row was bent at both ends (figure 4.32.1). The majority of femoral fractures in the middle section of the aircraft occurred to the occupant of the central seat of the row where the anterior cross member was more rigidly supported. Fractures of the femur may have occurred as a result of that femur being loaded over this spar. This hypothesis is supported by the increased incidence of femoral fractures in middle seat position (with a more rigid anterior spar), lack of soft tissue witness marks around the knee joint and an increased incidence of femoral fracture in those individuals with a greater than average buttock knee length (and therefore a longer lever). In the outer and inner seats in the row the anterior spar was bent at its junction with the leg support, this being more pronounced on the side adjacent to the fuselage. This bending would have to some extent have cushioned the transmission of forces to the femur.

Figure 4.32.1 Bending of the Anterior Spar in Passenger Triple Seat



The different femoral fracture types seen might also be related to this loading of the femur over the anterior bar. For example the occupant seated in 11A demonstrated bilateral femoral fractures. The types of fracture have been analysed using the the A.O. classification of fractures. The right femoral fracture was a short spiral and the left fracture had a large lateral butterfly. The anterior spar demonstrated a bend. It can be postulated that with loading of the femur over the anterior spar, the spar failed and bent. The effect of the spar bending would have the effect of twisting the lower limbs towards the fuselage and this has been confirmed by the passengers own statement. This resulted in a torsional injury to the right femur and a bending injury to the left femur. Significant axial loading of the femur did not occur as demonstrated by the occupant's lack of soft tissue injury around the knee.

The only association of the position adopted at the time of impact with the injuries sustained is an increased incidence of femoral fracture occurring in the 'braced' position. This may possibly reflect a higher bending moment created around the femur as a result of adopting a braced position. In the upright (unbraced) position the loading of the femur may be decreased because of the flailing of the upper limbs, and torso into the seat in front.

4.4 Summary

A wide variety of pelvic and lower limb injuries have been sustained by passengers and crew. In those areas of the aircraft that remained intact the occupants sustained severe injuries as a result of the primary forces and interactions with their surroundings. The range of injuries seen cannot be wholly explained in terms of individual variation (ie age, sex, weight, height etc.). Mechanisms other than the knee-femur-pelvis complex (axial loading) are apparent in the causation of femoral fractures and perhaps pelvic dislocations and fractures. Destruction of the floor in the forward and aft sections of the aircraft resulted in numerous distal tibial, ankle and foot injuries. Flailing of the lower limbs under the seat in front accounts for some of these tibial, ankle and foot fractures. Fractures to the tibial plateau and some ankle fractures are a result of axial loads indicating that flailing did not occur.

Chapter 5

Impact Testing

5.1 Introduction

The primary function of any occupant protection system is to protect the occupant from sustaining severe or fatal injuries. Review of the pelvic and lower limb injuries has revealed a wide variation in the types of injuries identified, for those passengers seated in the mid section of the aircraft (rows 10-20). The injuries were sustained despite the integrity of the seating and restraint system and the maintenance of the structure. Mechanisms of causation have been suggested with no unifying mechanism applying to all individuals. It is apparent that factors other than individual variation may have been important in modifying the impact biomechanics and kinematics of the occupants following impact.

The purpose of the following series of experiments, using a deceleration sled facility and anthropomorphic test device's (ATD), was to further assess and define the biomechanics of the pelvis and lower limb injuries sustained by the occupants seated in the mid section (rows 10-20) of the Boeing 737-400 aircraft (G-OBME). The protocol aims to investigate the effect of the position adopted at the time of impact and the effect of differing lap belt tensions on the occupant impact biomechanics and kinematics.

5.2 Experimental Design, Materials and Methods

The experimental design follows guide lines suggested by RF Chandler (1985 and 1987) and the FAA Advisory Circular 1988.

Dynamic test facility

The deceleration sled facility at the Royal Air Force Institute of Aviation Medicine, Farnborough, Hampshire was used. A full description of the facility was described by AF Giles in 1971 and AD Dutton in 1974.

The facility consists of a wheeled vehicle which runs on a track of 46 metres in length. Vehicle propulsion is provided by the energy stored in stretched bungee cords attached to a pusher or bogey. The number of bungee cords can be altered thus affecting the amount of stored energy. The system is primed by winching back the test vehicle and pusher. On release the vehicle accelerates over a distance of 26m followed by a coast phase of 13m. Deceleration is achieved by a steel cable harness stretched across the track, each end being connected to the piston of a hydraulic cylinder. On impact the hydraulic fluid in the rams are forced through a pair of orifices whose dimensions can be adjusted. By control of the variables a deceleration pulse of approximately half sine wave shape, with a variable peak G and duration, may be achieved.

Test fixture

FAA advisory circular states (1988) that head and knee impact conditions are best evaluated through a multiple row test fixture. Undamaged or minimally damaged seat rows were taken from G-OBME with permission of the insurers Lloyds of London. The test fixture was designed to simulate the interior of the G-OBME as closely as possible.

Two left hand rows of seats were mounted to the test vehicle at a designated 32 inch (81.3cm) seat pitch. A floor was constructed and carpeted on which to place the ATDs' feet. Panelling was removed from the arm rest of the outside seats in order to facilitate recording of displacement data. Plasticine to the depth of 0.8 cm was placed along the posterior longitudinal spar and the back of the knee panel of the seat in front. This was to assess contact of the test dummy with the seat in front. The seats were orientated in the Gx plane to simulate -Gx horizontal impacts.

Anthropomorphic test device

An instrumented 50th percentile hybrid III ATD was used as the experimental model. Anthropometric measurements made on the occupants seated in the mid section of the aircraft suggested that the 50th percentile hybrid III anthropomorphic test device was a good surrogate as it was of a similar size to that of the average anthropometric measurements taken on the occupants seated in this section (reviewed Appendix 3). The hybrid III dummy was placed in the rear row of the two row configuration, and in the outside seat to enable data collection. An OPAT dummy or Sierra Sam dummy was placed in the outside seat in the forward row of seats in front of the hybrid III ATD, in order to create the correct response and contact environment for the experimental model seated behind. This was placed in a recognised crash brace position for all tests.

The instrumented Hybrid III ATD was dressed in the following:

a) Cotton shirt

b) Polyester and cotton trousers, no belt.

c) Hushpuppy foam soled shoes

Instrumentation

Table 5.2.1 lists the recording devices used to collect data and figure 5.2.2 outlines the overall set up of recording equipment.

The vehicle was fitted with an accelerometer to record the acceleration of the test fixture. Further accelerometers were placed in the pelvis of the experimental model to measure pelvic Gx and Gz accelerations. A head accelerometer was used to provide additional information to help validate a computer simulation. Lap belt loads were measured with a pre-calibrated force link attached in series to the lap belt. This had the effect of decreasing the amount of webbing in the belt by 14cm. Knee shear gauges were located in the knee assembles of both left and right knees.

Table 5.2.1

Transducer	Description manufacturer	Type	Serial number	Range	Sens
Vehicle G	Strain gauge accelerometer (Pioden)	UA1	1071	± 50g	3.7 mV/G
Pelvis Gx	Strain gauge accelerometer (J.P.B.)	J505-50-F3	3070	± 50g	1.3 mV/G
Pelvis Gz	Strain gauge accelerometer (J.P.B.)	J505-50-F3	3071	± 50g	1.3 mV/G
Head Gx	Strain gauge accelerometer (Kyowa)	PR 93 7/50	L0701	± 50g	58μV/G
Left knee	Sliding knee potentiometer (Carter MFG.Co.)	FCPST-632 711-1	115091	-	-
Right knee	Sliding knee potentiometer (Carter MFG.Co.)	FCPST-632 711-1	115091	-	-
Lap belt	Quartz force link (Kistler)	9321A	300793	± 10 kN	4.02pc/N

Instruments were connected to datalab 2000 transient recorder, with a sample time of 5,000 per second, by means of fly leads. The raw data was then either recorded to a Gould ES 1000 plotter, which allowed measurements to be taken directly from paper records, or transferred directly into a computer data base.

Overall Set-up of Recording Equipment

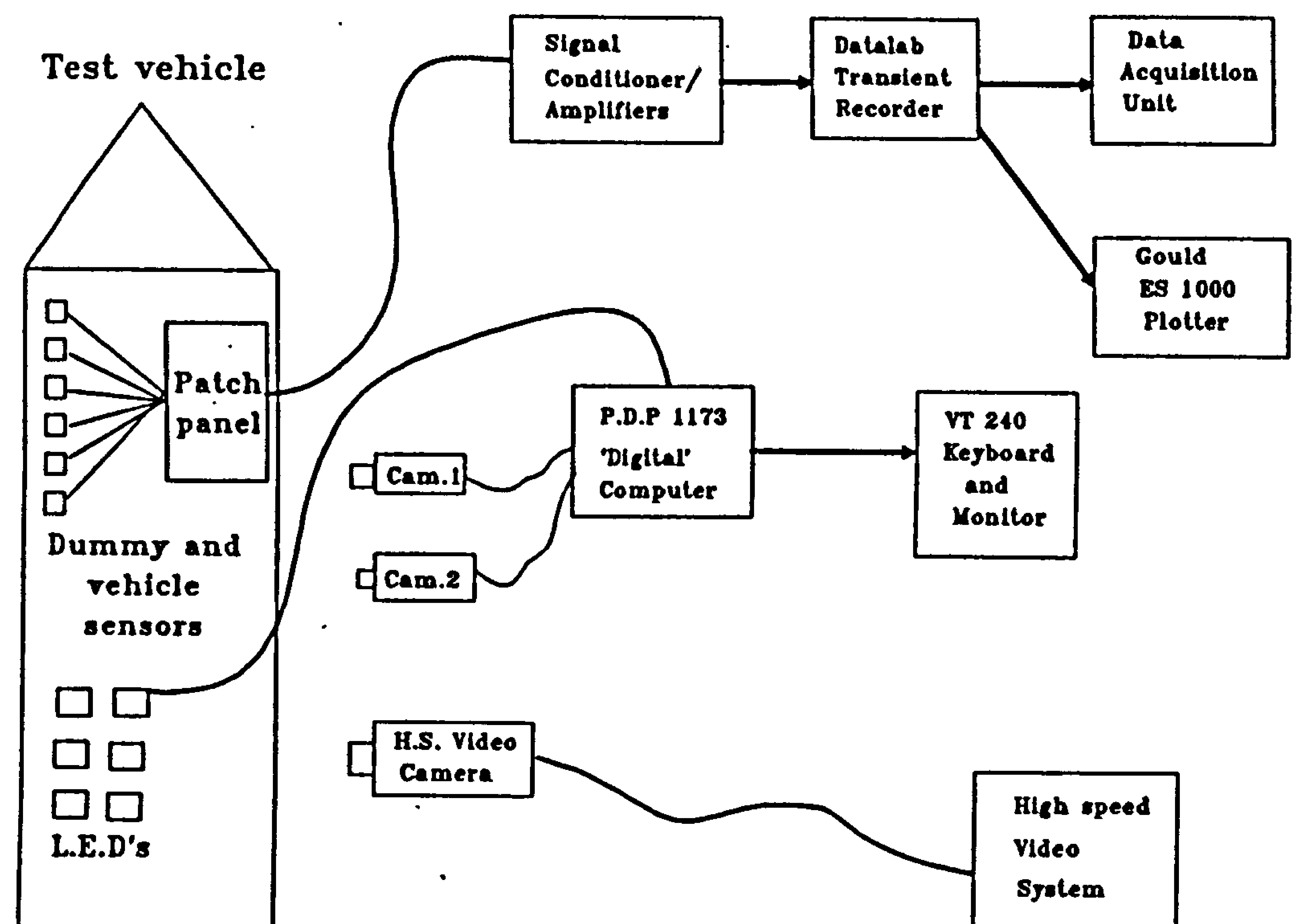


Figure 5.2.2

Displacement (trajectories) data for the pelvis, thigh and ankle was achieved with the use of the Selspot Motion Analysis System which offers a means by which the space time co-ordinates of specific points on a moving system can be recorded. At the points of interest infrared light emitting diodes were attached. These diodes flash in sequence and their output was detected by two cameras and analysed by computer giving a three dimensional localisation of any point. A more complete description of the Selspot Motion analysis system is given by McKenzie (1987).

Placement of the light emitting diodes (LED) were as follows (see figures 5.2.4 -7):

a) Reference LED, to the structure of the rear seat row.

b) Pelvic LED, the highest point overlying the iliac crest, approximately 17cm from seat squab.

c) Thigh LED, 22cm from the patella anterior surface (knee flexed to 90 degrees). The placement of this LED was such so as not to interfere, with the knee assemble of the ATD.

d) Ankle LED, over lying the lateral malleolus of the dummy, 12 cm from the floor (no shoes on dummy).

The output of the Selspot motion analysis system can be made available in graphic form. Measurements of displacements are then taken directly from this output.

All impact tests were recorded on a high speed video camera system, recording at 40 frames per second. This allowed for immediate viewing of the test impact confirming the correct positioning of the lap belt and ATD prior to impact. It also recorded the trajectory and general kinematics of the dummy.

Calibration

The accelerometers were calibrated on a centrifuge of known radius. Revolutions per minute were recorded on a calibrated tachometer. Calibration signals were therefore obtained by connecting the accelerometer to the datalab whilst centrifuging at a known rate. Calibration was carried out at the beginning of the experiments and at the end to confirm the integrity of the system.

The lap belt force link was pre-calibrated at manufacture. A calibration signal was injected into the datalab and provided a signal that allowed conversion to engineering units.

Knee shear gauge calibration data is reviewed in Appendix 4.

The Selspot Motion Analysis system was calibrated at the beginning of every days testing, using the suppliers position reference structure.

Experimental conditions

The impact of the Boeing 737-400 (G-OBME) resulted in two main acceleration vectors. The nature of the accelerations involved in the mid section of the aircraft have been reviewed in Chapter Three and by Cranfield Impact Centre (Sadeghi et al 1989). An initial horizontal component (-Gx) of 15-20G lasting approximately 100 ms was followed by a vertical component (+Gz) occurring after the initial -Gx acceleration. Unfortunately present systems of dynamic crash test facilities are unable to simulate a multi-directional acceleration crash pulse or the velocities (Chandler 1971, 1987) and therefore the energy involved in a crash. Dynamic test facilities can however simulate the magnitude of the acceleration change. A horizontal acceleration vector (Gx) has therefore been used in the experimental design at three designated levels 9G, 16G and 20G for a duration of 80-100ms. Table 5.2.3 records the velocities and pulse durations at the three designated G employed in the experiment.

Table 5.2.3

Designated G	Vehicle Velocity (M/s)	Pulse duration (ms)
9	4.95 ± 0.1	95 ± 5
16	6.6 ± 0.1	85 ± 5
20	7.82 ± 0.05	80 ± 5

Positioning of dummy

The effect of positioning of the lower limbs and brace position adopted by the ATD were investigated. Figure 5.2.4 (braced -feet forward), 5.2.5 (braced -feet back), 5.2.6 (unbraced -feet forward) and 5.2.7 (unbraced -feet back) demonstrates the positions investigated. The ATD was placed in the centre of the outside seat with the back and buttocks against the seat back. In the braced position the dummy was bent forward until the head was in contact with the seat in front. The ATD was sat upright in the seat in the unbraced position. For braced positions the arms were taped behind the head, and for the unbraced positions crossed in front of the abdomen. Two lower limb position were identified: Feet forward, ie. at an angle of 20 degrees to the vertical. This position represented a position in which the legs could be easily placed when the dummy was in an unbraced position; Feet back, ie. an angle of 12 degrees to the vertical. In this position the feet of the dummy were planted on the floor. The dummy's knees were separated by four inches (102mm) and were approximately eight inches (203mm) from the seat infront. Both positions represented a comfortable seating position.

Lap belt tension

Lap type seat belts were placed below the anterior superior iliac spines passing over the region of the greater trochanter of the femur. Two belt tensions were investigated in all experimental conditions. Tensions were measured using a spring balance as i) 20 pounds and ii) 40 pounds. Twenty pounds of tension represented a tension that

Figure 5.2.4

Positioning of Dummy
Braced - feet forward

Figure 5.2.5

Positioning of Dummy
Braced – feet back

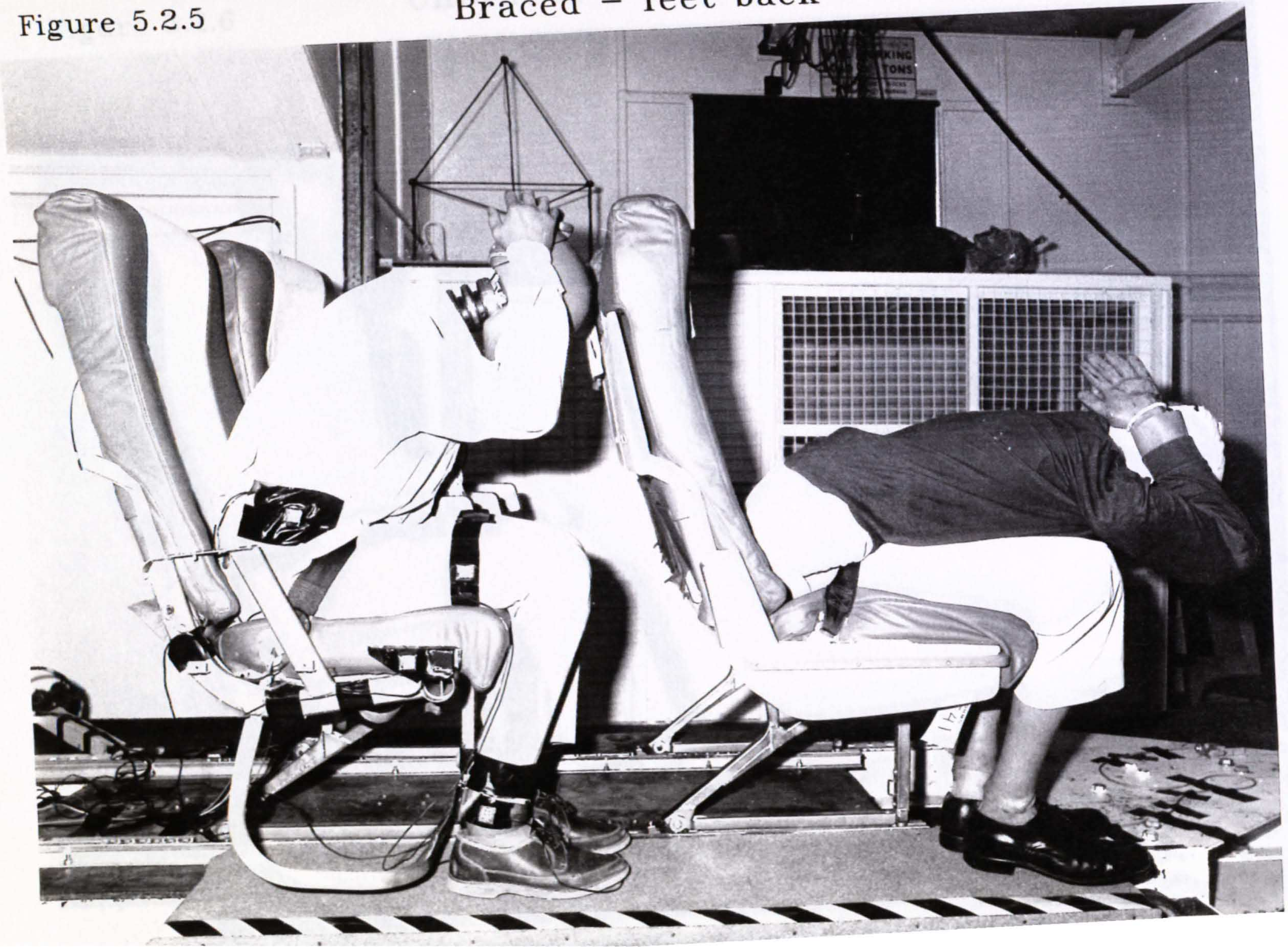


Figure 5.2.6

Positioning of Dummy
Unbraced - feet forward

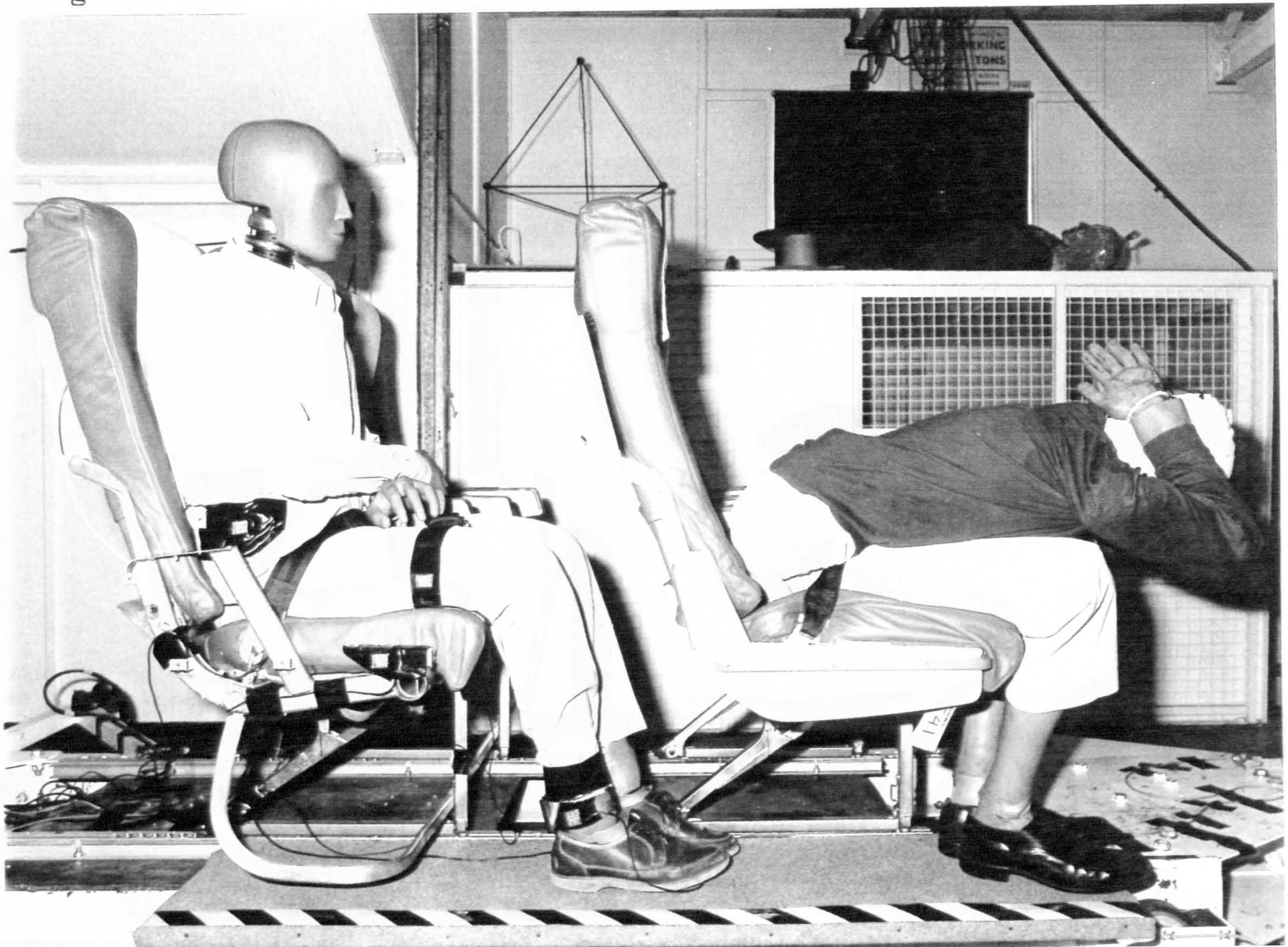
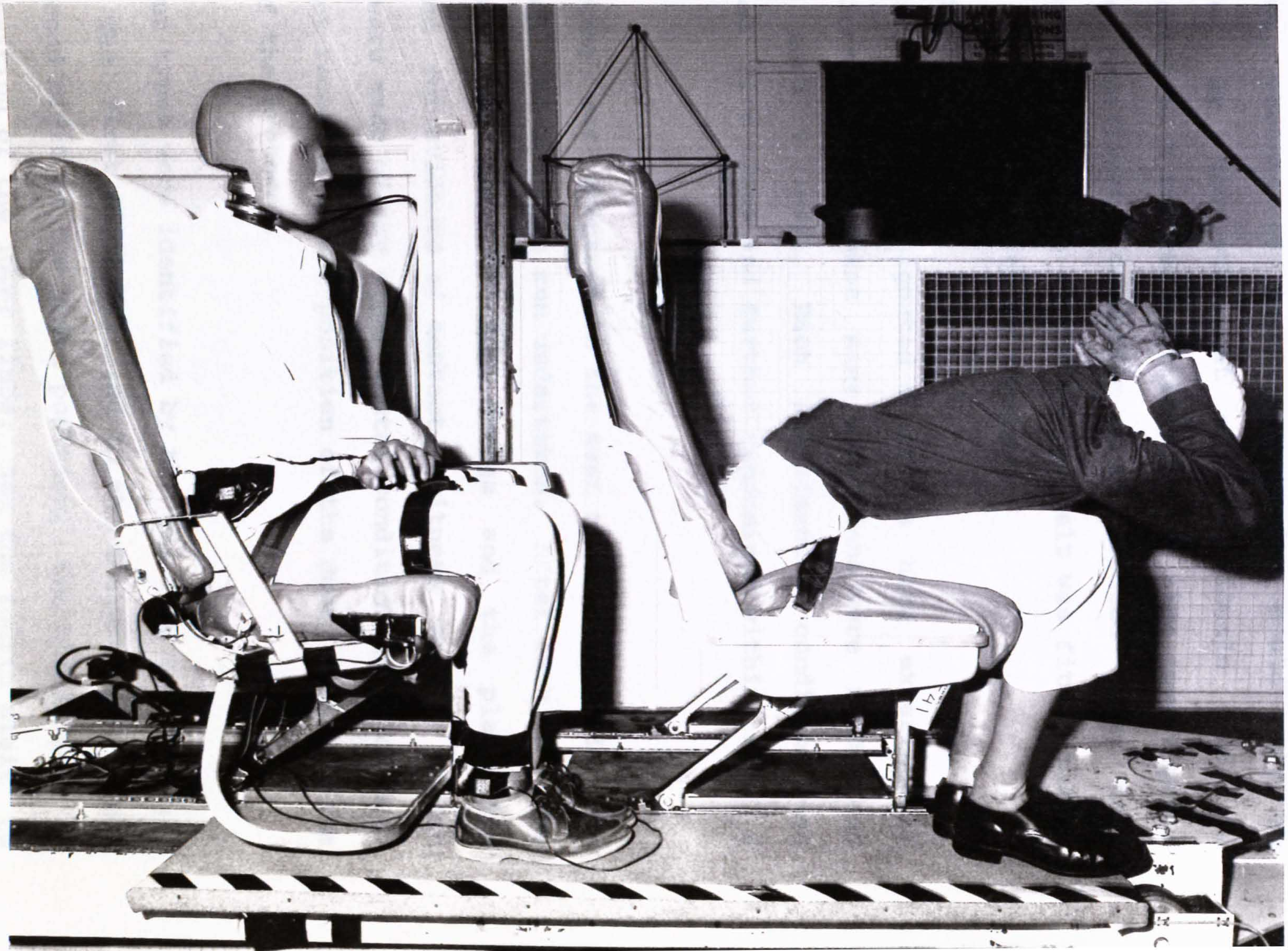


Figure 5.2.7 Positioning of Dummy Unbraced – feet back



would be easy to apply and would be comfortable for an occupant, whereas 40 pounds of tension could only be achieved by the strongest of individuals and is uncomfortable. A further belt tension was also investigated in the experimental condition of an unbraced dummy with feet forwards. In this situation the belt was fitted snugly to the occupant but with no tension.

Procedure

For each of the designated G levels nine experimental conditions exist. These nine conditions were randomised within each G level. Each experimental condition was repeated five times and further randomised within each G level.

The dummy was prepared in the seat as described in the protocol and the test run undertaken. After each run the equipment was examined for damage and the plasticine examined for evidence of contact (witness mark). Repeat tests were made of any experimental condition in which data was not recorded or the position of the dummy altered from that of the protocol.

The run types were identified by the use of a four figure code. The first digit referred to the designated G level; the second digit to the brace position; the third digit to the position of the lower limbs; and the fourth digit to the belt tightness. Table 5.2.8 contains a summary of the run codes and types. Thus run code of 2122 refers to a 16G, unbraced, legs forward, 40lbs belt tension run.

Table 5.2.8

Digit	Parameter	Code 1	Code 2	Code 3
First	G level	9G	16G	20G
Second	Brace position	Unbraced	Braced	-
Third	Leg position	Back	Forwards	-
Fourth	Belt tension	20lbs	40lbs	0lbs

Experimental measurements

The following measurements (table 5.2.9) were recorded for each experimental condition.

Table 5.2.9

Parameter Measured	Abbreviation	Units
Vehicle impact pulse.	V G	G
Pelvic Gx impact pulse	P Gx	G
Pelvic Gz impact pulse	P Gz	G
Lap belt load	STRap	kN (Kilonewtons)
Left leg shear	LLeg	N (Newtons)
Right leg shear	RLeg	N (Newtons)
Pelvic X displacement	Pdis X	mm (millimetres)
Pelvic Z displacement	Pdis Z	mm (millimetres)
Thigh X displacement	Kdis X	mm (millimetres)
Thigh Z displacement	Kdis Z	mm (millimetres)
Ankle X displacement	Adis X	mm (millimetres)
Ankle Z displacement	Adis Z	mm (millimetres)

where x = horizontal displacement
z = vertical displacement

The left knee potentiometer unfortunately had a fault and was therefore excluded from the analysis of the results (as explained in Appendix 4).

Analysis of results

Raw data was entered into a computer data base. The raw data was manipulated using software developed by the Institute of Aviation Medicine, Farnborough. Further software was developed in order to further analyse the Pelvic Gz component (vertical accelerations) recorded by the pelvic accelerometer. Recordings of the runs from the * Gould plotter for the pelvic Gz parameter revealed a bi-phasic vector response in some of the test situations. The software developed recorded the maximum footwards acceleration (PGz pos), and the maximum headwards acceleration (PGz neg), for each test run. The maximum resultant pelvic acceleration (PG res) was calculated as being the greatest acceleration (G) at any given time using the formula PG res = $\sqrt{(PGzm^2 + PGx^2)}$, where PGzm is the maximum acceleration in the z (vertical) plane and PGx the acceleration in the x (horizontal) plane.

All results were entered onto a spread sheet allowing some simple manipulation of the data. Calibration data for the load cells was incorporated to express data in the correct units.

Statistical analysis was undertaken using software, developed by the Institute of Aviation Medicine, Farnborough, as an analysis of variance (ANOVA). Four elements corresponding to the controlled parameters were identified, brace position, leg position, seat belt tightness, and designated 'G' level. The effects of the controlled parameters were investigated using multi-way contingency tables. A fifth non controlled parameter was identified and included on the basis of whether the legs flailed or not.

The assumptions of ANOVA - normality, addivity and homogeneity of variance - were checked by determining the best power transformation to ensure conformity with the assumptions, using the maximum likelihood method of Box and Cox. A transformation was selected if it was strongly suggested on the basis of the appropriate chi-square test.

5.3 Results

The results will be considered in 3 sections, motion of the lower limbs as seen by video recordings, Selspot displacement data, and datalab recorded data.

Tables 5.32.1 and 5.33.1 summarize results for each run type, as means and standard deviations. In Appendix 5 the results from all experiments are recorded.

5.31 Motion of the anthropomorphic dummy

The behaviour and trajectory of the the ATD seated in the rear row of the test fixture were recorded by a high speed video recorder. From the review of the recordings it was apparent that lower limb kinematics could be divided into two groups :- a) those in which the legs flailed, ie. the lower limbs flailed in a smooth arc under the seat in front with the shin or ankle striking the posterior lateral spar and the feet striking the sub seat assemblies. This was confirmed by indentations in the plasticine placed along the posterior spar; b) those in which the legs did not flail under the seat in front. In the experimental conditions in which the limbs did not flail no witness mark was seen in the plasticine placed along the posterior spar.

Flail group

Experimental runs that demonstrated flail behaviour, in all test replications, were those in which the ATD was placed in an unbraced position with the legs in a forward position. The lap belt tension did not affect the flail behaviour. The flail behaviour of the lower limbs is demonstrated in series 5.31.1. It can be seen that flailing of the lower limbs forces the knee into an extended position.

Series 5.31.1

Flail Behaviour

as seen in run 3122

Series 5.31.2

Non Flail Behaviour

as seen in run 3111 (unbraced)

Series 5.31.3

Non Flail Behaviour

as seen in run 3212 (braced)

The run types that demonstrated flail behaviour were:-

1121	1122
2121	2122
3121	3122 (3123)

Non flail group

Absence of flailing was seen in test conditions where the lower legs were placed in a legs back position. Motion of the lower limbs, as recorded on high speed video recorder, is demonstrated in series 5.31.2 in the unbraced position and 5.31.3 in the braced position. As can be seen the feet slide forward on impact but do not flail in front of a vertical line drawn through the knee.

The run types that demonstrated an absence of flailing were:-

1111	1112	1211	1212
2111	2112	2211	2212
3111	3112	3211	3212

Flailing of the lower legs was also not seen in the braced position with the feet forwards at 9 G or 16 G. In these test runs the feet were seen to slide forward on impact with a stuttering motion, some test runs demonstrating a slight 'flip' at the end of the excursion (ie. foot just lifting of the floor), but insufficient to impact with the seat in front.

Run types that demonstrated this behaviour were:-

1221 1222

2221 2222

At 20 G a variable response was seen with some impact tests resulting in flailing of the lower limbs with impact against the plasticine marker on the posterior spar of the seat (five cases), and in others the stuttering behaviour described above (five cases). Video recordings of those cases that flailed revealed a violent flailing of the lower limb under the seat in front with some hyper-extension of the knee. The run types that demonstrated this flail behaviour in some of the test runs were:-

3221 3222

5.32 Selspot displacement data

This section reviews displacement data, relating to the pelvis, knee and ankle, as measured by Selspot Motion Analysis System. The results for each run type are expressed as a mean (ave) and standard deviation (std) and are recorded in table 5.32.1. Displacements are measured in millimetres (mm). The abbreviation dX refers to motion in a horizontal plane and dZ motion in a vertical plane.

i) Ankle displacement

From table 5.32.1 it can be seen the average horizontal (A dX) and vertical (A dZ) ankle displacement are recorded. Maximum ankle horizontal and vertical motion occurred in

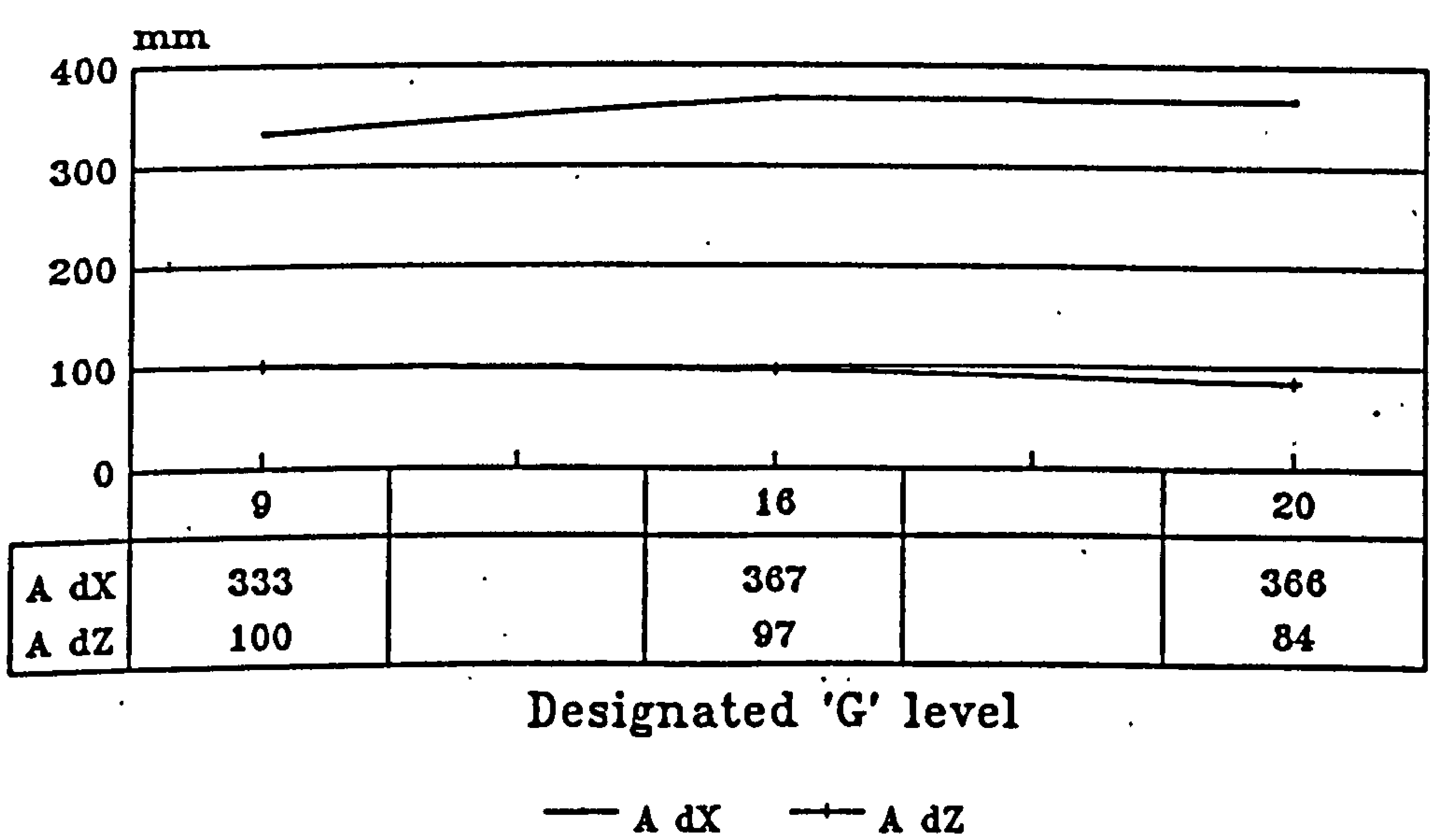
Selspot Displacement Results

TYPE	P dX ave	P dX std	P dZ ave	P dZ std	Kd X ave	K dX std	K dZ ave	K dZ std	A dX ave	A dX std	A dZ ave	A dZ std
	(mm)		(mm)		(mm)		(mm)		(mm)		(mm)	
1111	175.00	11.73	10.40	1.08	135.20	12.01	9.50	1.66	145.60	19.02	12.50	3.71
1112	169.40	11.44	11.70	4.84	120.60	14.38	10.20	2.20	163.60	29.58	12.20	1.15
1121	194.40	13.39	15.30	1.82	136.00	10.84	57.60	7.83	333.00	18.91	83.00	24.14
1122	165.00	16.20	12.30	1.20	112.80	12.09	43.60	5.59	324.00	8.94	100.00	7.91
1123	197.80	5.12	21.60	2.49	162.00	5.87	36.50	19.18	209.00	45.40	17.80	5.06
1211	170.60	16.10	20.20	6.50	139.80	12.26	21.40	2.38	93.20	9.15	9.10	1.52
1212	146.40	5.46	21.10	5.27	118.40	7.44	15.80	2.39	81.20	8.38	5.40	1.14
1221	167.00	11.51	18.30	2.22	140.00	9.35	11.80	4.71	118.00	25.39	7.70	2.28
1222	140.20	5.45	17.10	2.97	113.80	7.19	28.80	14.74	194.60	95.79	46.10	41.16
2111	198.80	5.07	27.30	5.19	158.40	5.94	12.80	4.31	169.80	17.01	21.70	13.10
2112	182.00	7.58	23.80	3.96	142.80	6.10	12.90	3.31	190.80	21.43	19.50	4.47
2121	221.80	10.06	30.90	5.41	168.20	8.43	89.80	16.39	313.80	60.30	70.80	29.39
2122	198.00	3.54	26.00	5.10	145.80	6.76	85.20	8.11	332.40	19.53	97.40	12.86
2123	227.20	7.66	33.20	3.70	178.20	13.05	112.80	2.17	367.60	12.40	88.60	21.96
2211	195.60	16.18	34.60	6.77	159.60	16.82	26.50	2.40	125.60	12.74	18.80	3.47
2212	183.80	11.14	32.20	4.66	145.20	12.78	24.30	1.57	120.20	15.90	17.10	3.86
2221	193.80	18.51	36.60	4.10	156.60	17.36	39.70	21.70	188.60	44.63	16.00	2.74
2222	182.20	15.06	34.00	4.53	147.60	8.79	48.60	24.25	195.00	55.92	18.40	2.97
3111	223.60	11.39	36.00	2.35	186.40	10.90	45.10	26.10	279.40	65.66	26.60	5.22
3112	215.20	2.89	35.90	3.58	173.20	3.70	11.70	4.12	223.00	17.28	23.00	3.67
3121	226.60	3.85	37.00	2.55	189.20	9.88	126.20	4.44	365.60	16.62	84.60	22.90
3122	216.40	9.10	36.30	2.49	174.80	11.97	115.00	5.48	366.60	14.52	81.20	10.33
3123	229.40	6.69	36.00	2.24	191.20	4.44	121.00	7.59	353.40	14.74	80.80	23.47
3211	206.60	9.99	40.80	3.96	174.80	8.56	22.90	2.92	178.80	47.20	16.70	2.08
3212	210.40	9.79	43.40	4.45	175.40	9.63	24.70	2.54	157.40	5.73	16.90	1.29
3221	220.60	8.62	41.80	4.32	189.40	8.47	92.40	31.34	282.80	40.68	37.80	23.97
3222	201.60	5.32	43.50	2.69	165.00	8.63	103.60	26.82	294.60	35.68	56.20	20.67

Table 5.32.1

test runs with the ATD unbraced and the feet in a forward position, whereas minimum motion was seen when the dummy was braced and the legs placed in the posterior position. Figure 5.32.2 plots the maximum ankle displacement (horizontal and vertical) at each designated G level. From this graph it can be seen that horizontal ankle displacement increases with 'G', tending to plateau at 20G, indicating a maximum excursion is reached. Conversely with increasing 'G' ankle vertical displacement decreases.

Maximum ankle displacements at each 'G'
(A dX and A dZ millimeters)
(for all test conditions)



	9		16		20
A dX	333		367		366
A dZ	100		97		84

Figure 5.32.2

Using Days method of clustering on ankle displacement measurements, A dX and A dZ, two groups can be identified. One group in which the ankles appear to displace a greater distance, this represents flailing of the lower limbs, and the other group where displacement is less, this represents a group where no flailing of the lower limbs has occurred. From this analysis 35 flails can be identified out of 120 trials. The allocation produced by this method was checked against the video analysis and found to correlate to those cases identified as flail or non flail according to the motion seen. Thus flailing was associated with both leg position ($p=<0.0001$) and bracing ($p=<0.0001$). On no occasions when the legs were back did the legs flail. In no trials where the dummy was in the braced position at 9 G and 16 G runs did flailing occur, although on five occasions at 20 G flailing occurred. Flailing was associated with an unbraced 'upright posture' with the feet forward.

Flailing behaviour was therefore included as a fifth uncontrolled factor in the analysis of variance as having an additional effect. Flail or non flail behaviour was identified as being an uncontrolled variable and identified on video data and ankle displacement recordings. Flail behaviour is related to brace position ($p=<0.0001$) and limb position ($p=<0.00010$) as previously described. It is also associated with bracing and 'G' level ($p=<0.005$). Eighty five test conditions resulted in flail behaviour and 35 conditions did not. For the analysis of variance the effect of flail has been considered in the analysis as an uncontrolled variable.

Analysis of variance for horizontal ankle displacement

The most important effect on horizontal ankle displacement (A dX) is the effect caused by flailing motion. Ankle displacement in the horizontal plane can also be seen to vary with the following parameters in the way indicated in the table 5.32.3.

Table 5.32.3

Source of variation	Significance (p=<)	Effect on horizontal ankle displacement
Brace position	0.001	Increase with unbraced
Leg position	0.001	Increase with forward
'G' level	0.001	Increase with G
Flail	0.001	Increase with flail

Analysis of variance for vertical ankle displacement

Ankle displacement in the vertical plane (A dZ) was found to vary with the following parameters (table 5.32.4). Again the most important effect was that of flailing conditions.

Table 5.32.4

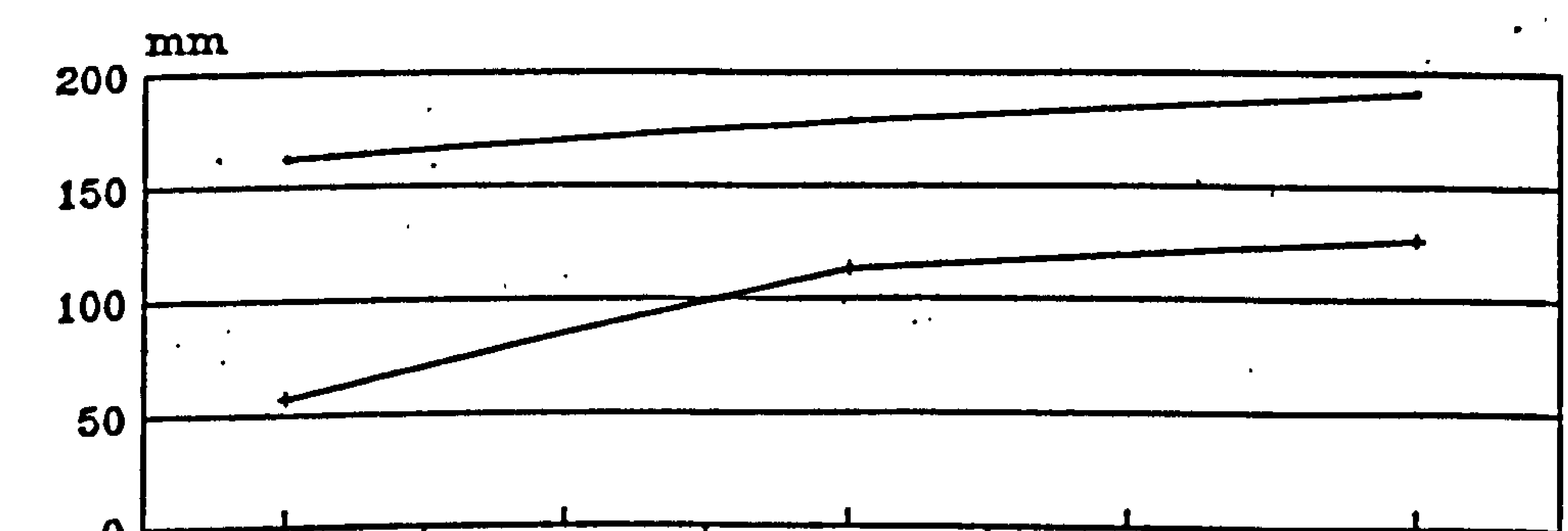
Source of variation	Significance (p=<)	Effect on vertical ankle displacement
Brace position	0.001	Increase with unbraced
Leg position	0.01	Increase with forward
'G' level	0.001	Increase with G

Leg and strap	0.001	Increase back < tension Increase forward > tension
Leg and 'G'	0.001	Increase forward + G Increase back + G
Flail	0.001	Increase with flail

ii) Knee displacement

Knee displacement results (K dX and K dZ) are demonstrated in table 5.32.1. The maximum horizontal (K dX) and vertical excursion (K dZ) of the thigh LED (knee) at each designated G is plotted in figure 5.32.5. Maximum horizontal knee displacement (K dX) can be seen to increase with G. Vertical displacement (K dZ) increases with designated G level, but appears to plateau at high G.

Maximum knee displacements at each 'G'
(K dX and K dZ millimeters)
(for all test conditions)



	9		16		20
K dX	162		178		191
K dZ	57		113		126

Designated 'G' level

— K dX —+— K dZ

Figure 5.32.5

Maximum Kdis X was seen to increase in a linear manner with 'G'. Examination of the plasticine witness indicator, placed on the knee panel of the seat in front, indicated for the test conditions investigated no indentations. It can thus be concluded impact to this region of the seat did not occur. A further experimental variable was therefore introduced, that of a seat belt tension of 0 lb., for the unbraced, feet forward test condition. This was investigated at each of the designated 'G' levels (ie runs 1123, 2123 and 3123). These conditions were considered to be optimal to investigate contact of the knee with the back of the seat in front. Only at 20G was minor indentation seen in the plasticine. Review of the video film also indicated that knee contact had not occurred, with the knee panel of the seat in front.

Analysis of variance for horizontal knee displacement

Horizontal knee displacement varied with only two parameters (table 5.32.6).

Table 5.32.6

Source of variation	Significance (p=<)	Effect on horizontal knee displacement
Belt tension	0.001	Decrease with > tension
'G' level	0.001	Increase with G

Analysis of variance for vertical knee displacement

Knee displacement in the vertical plane was represented as a down wards displacement. Displacement was found to vary with the following parameters as indicated in the table 5.32.7 below. Greatest displacement is seen in those situations when the lower legs are in a forward position.

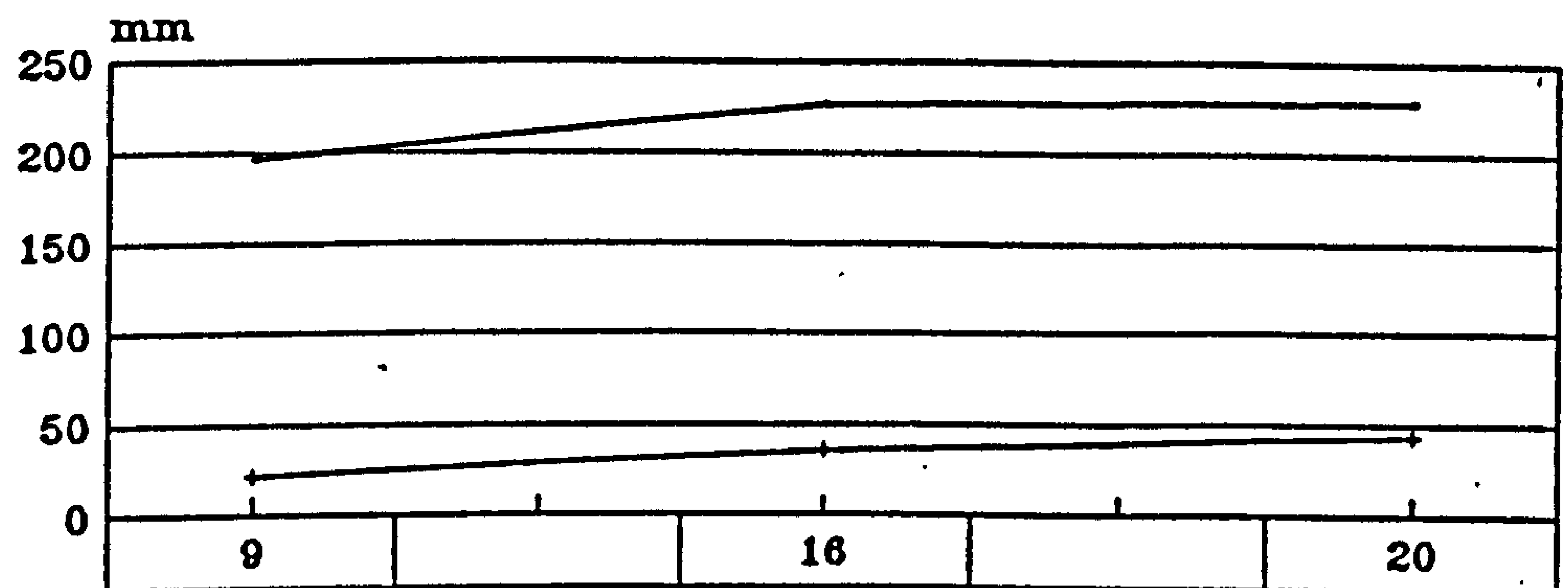
Table 5.32.7

Source of variation	Significance (p=<)		Effect on vertical knee displacement
Leg position	0.001		Increase with forward
'G' level	0.001		Increase with G
Brace and leg	0.001		Increase unbraced forward Increase braced forward
Leg and 'G'	0.001		Increase forward + G Increase back + G
Brace, leg and 'G'	0.01	9G	Increase unbraced forward Decrease braced forward
		16G	Increase unbraced forward Increase braced forward
		20G	Increase unbraced forward Increase braced forward
Leg and strap	0.05		Increase back < tension Increase foreward > tension
Flail	0.05		Increase with flail

iii) Pelvic displacement

The means and standard deviations of the displacement data for horizontal (P dX) and vertical (P dZ) pelvic motion are recorded in table 5.32.1 for each test condition. Figure 5.32.8 plots maximum pelvic displacements at each designated 'G' level. It can be seen that horizontal pelvic displacement appears to plateau at high 'G' levels whereas

Maximum pelvis displacements at each 'G'
(P dX and P dZ millimeters)
(for all test conditions)



	9		16		20
P dX	197		227		229
P dZ	22		36		43.5

Designated 'G' level

Figure 5.32.8

with 'G'. In all test situations horizontal pelvic motion, as measured from the anterior superior iliac spine, was seen to be of greater magnitude than horizontal knee displacement. No association was seen with flail behaviour.

Analysis of variance for horizontal pelvic displacement

Table 5.32.9 records the important causes of variance for horizontal pelvic displacement.

Table 5.32.9

Source of variation	Significance (p=<)	Effect on horizontal pelvic displacement
Brace position	0.001	Increase with unbraced
Belt tension	0.001	Decrease with > tension
'G' level	0.001	Increase with G
Leg and strap	0.05	Increase back < tension Increase forward > tension

Analysis of variance for pelvic vertical displacement

Pelvic displacement in the vertical plane has been found to vary with the following parameters in the way indicated in table 5.32.10.

Table 5.32.10

Source of variation	Significance (p=<)	Effect on vertical pelvic displacement
Brace position	0.001	Increase with braced
'G' level	0.001	Increase with G

Analysis of variance of displacement parameters

Analysis of variance was carried out to investigate the relationship of the relative motions of the pelvis, knee and ankle in the two situations of flail (n=35) and non flail (n=85).

In the non flail situations, knee displacement in the horizontal plane was significantly (p=<0.001) affected by pelvic motion both in the horizontal and the vertical planes. However vertical knee displacement was only affected (p=<0.001) by pelvic displacement in the vertical plane.

Horizontal ankle displacement in the non flail situation was associated with both horizontal (p=<0.05) and with vertical (p=<0.001) knee motion. However vertical ankle displacement was just significantly (p=0.03) associated with vertical knee motion.

For the flail group (n=35) knee vertical and horizontal displacement varied with both pelvic horizontal and vertical motion (p=<0.001). Only horizontal knee displacement was found to be associated with horizontal ankle displacement (p=<0.05).

<u>5.33 Datalab recorded data</u>

The effect of the experimental variables on loads and accelerations experienced in the lower limbs and pelvis are discussed in this section. Raw data was recorded to a datalab 2000 transient recorder from accelerometers and strain gauges placed in the pelvis, knee assembly and lap belt. Table 5.33.1 records the means and standard deviations for each run category.

i) <u>Lap belt force link</u>

The lap belt force link loads (STR) for each run type, are recorded in table 5.33.1. The results are expressed as the

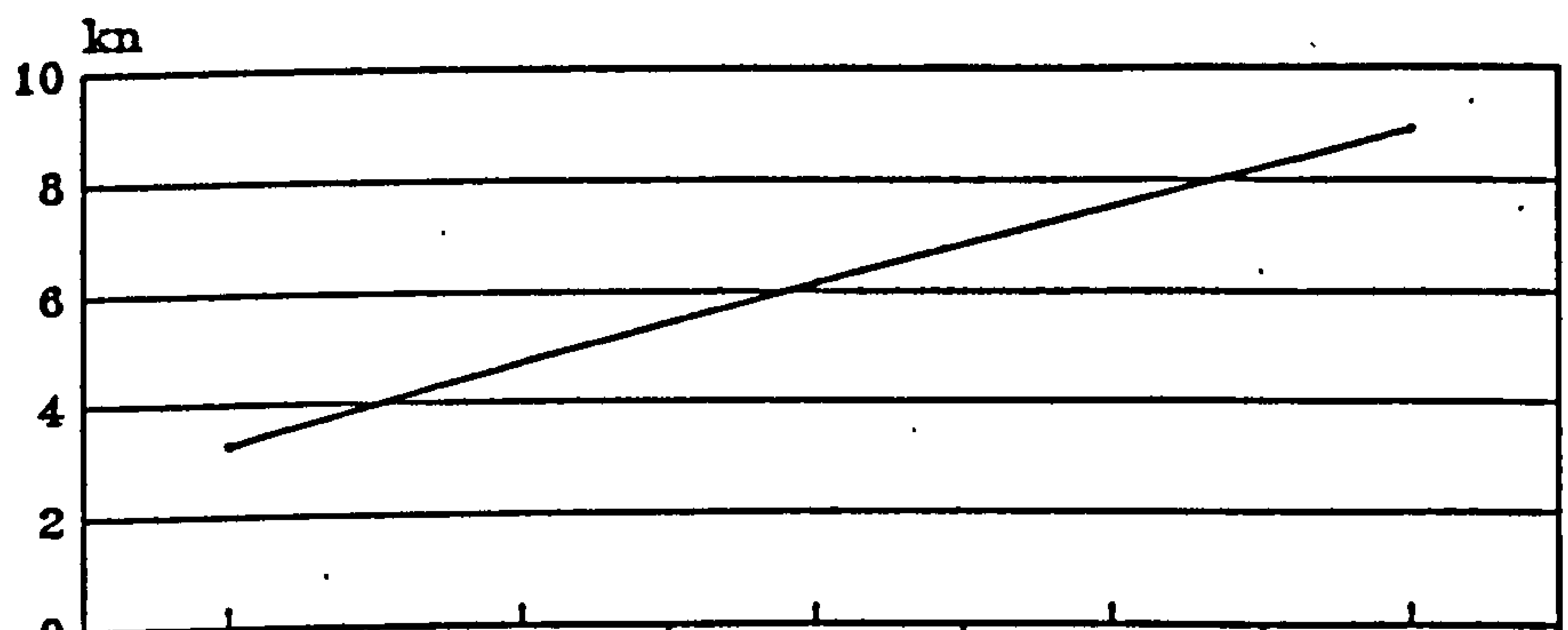
Datalab Recorded Results

TYPE	V G ave	V G std	PG ave	PGx std	PGzp ave	PGzp std	PGzn ave	PGzn std	PGzm ave	PGzm std	PGres ave	PGres std	STR ave	STR std	RL ave	RL std
	(G)		(G)		(G)		(G)		(G)		(G)		(kN)		(N)	
1111	8.89	0.11	9.79	0.41	3.11	0.31	2.26	0.74	3.19	0.21	9.87	0.40	2.71	0.09	43.68	24.20
1112	8.92	0.14	9.57	0.51	3.70	0.59	2.35	0.44	3.70	0.59	9.78	0.47	2.54	0.21	46.41	16.84
1121	9.08	0.20	10.07	0.60	3.80	0.54	2.37	0.39	3.80	0.54	10.13	0.58	2.63	0.19	96.46	44.16
1122	9.02	0.19	9.50	0.33	4.00	0.38	2.22	0.29	4.00	0.38	9.76	0.39	2.49	0.21	94.64	39.98
1123	8.95	0.16	10.50	0.76	3.33	0.39	2.59	0.45	3.35	0.39	10.44	0.79	2.72	0.16	41.86	60.57
1211	9.05	0.11	8.34	0.50	2.35	0.61	3.61	0.46	3.61	0.46	8.59	0.59	3.07	0.15	33.67	10.96
1212	8.93	0.37	8.37	0.94	2.15	0.35	3.33	0.66	3.33	0.66	8.43	0.95	3.20	0.37	36.40	8.51
1221	9.06	0.21	8.58	0.46	1.85	0.85	3.70	1.02	3.76	0.96	9.02	0.72	3.26	0.34	47.32	23.33
1222	8.97	0.26	8.86	0.67	2.00	0.51	3.41	1.11	3.41	1.11	8.91	0.65	3.24	0.36	35.49	18.31
2111	14.92	0.42	18.88	0.57	7.30	0.91	4.22	0.90	7.30	0.91	19.00	0.66	5.03	0.05	32.76	13.03
2112	14.93	0.22	18.50	0.56	8.06	0.83	4.30	0.59	8.06	0.83	18.72	0.67	4.92	0.10	30.94	16.21
2121	14.84	0.18	18.50	1.18	6.69	1.65	5.43	1.14	6.72	1.63	18.65	1.26	4.88	0.15	239.33	158.81
2122	14.88	0.16	18.64	0.70	8.22	0.58	4.81	0.68	8.22	0.58	18.70	0.50	4.90	0.26	172.90	79.46
2123	14.96	0.27	19.27	0.35	7.28	1.17	6.00	0.77	7.39	1.01	19.39	0.37	5.15	0.12	324.87	146.08
2211	14.98	0.18	16.02	0.33	3.07	0.23	5.54	0.58	5.54	0.58	15.94	0.30	5.93	0.17	35.49	9.86
2212	15.06	0.24	16.19	0.48	3.39	0.31	6.11	0.27	6.11	0.27	16.30	0.44	6.15	0.09	77.35	87.11
2221	14.82	0.40	15.96	0.87	2.76	0.61	5.04	0.47	5.04	0.47	15.93	0.79	5.58	0.16	46.41	25.21
2222	15.08	0.31	16.17	0.76	2.96	0.30	5.30	0.43	5.30	0.43	16.07	0.67	5.62	0.15	85.54	155.88
3111	20.23	0.22	26.50	0.42	10.02	1.02	5.17	0.48	10.02	1.02	26.85	0.42	7.77	0.27	20.93	4.07
3112	20.26	0.16	26.13	0.40	10.35	0.53	4.48	0.46	10.35	0.53	26.43	0.26	7.77	0.24	21.84	6.75
3121	19.94	0.58	26.77	0.44	10.13	2.68	6.72	1.17	10.22	2.54	27.31	0.50	7.73	0.31	281.19	57.95
3122	19.97	0.41	26.30	0.25	9.91	0.83	5.52	1.01	9.91	0.83	27.04	0.35	7.62	0.15	230.23	49.59
3123	20.11	0.19	26.77	0.62	8.72	2.09	5.26	1.46	8.72	2.09	27.04	0.77	8.02	0.28	415.87	203.45
3211	19.66	0.56	23.81	0.49	3.85	0.82	7.33	0.40	7.33	0.40	23.61	0.52	8.91	0.30	49.14	22.15
3212	20.28	0.23	23.57	0.87	4.67	0.49	7.57	0.55	7.57	0.55	23.46	0.72	8.79	0.19	55.51	11.78
3221	20.08	0.23	24.13	1.21	3.56	0.89	6.09	0.87	6.09	0.87	23.91	1.20	8.41	0.28	356.72	424.13
3222	20.09	0.31	24.39	0.29	4.31	0.57	6.44	0.30	6.44	0.30	24.17	0.29	8.34	0.17	697.06	390.82

Table 5.33.1

mean load in kilonewtons (kN) for each run type. The maximum lap belt load recorded for all run types at each designated 'G' are plotted graphically on figure 5.33.2. Lap belt loads were seen to increase with designated 'G' to a maximum of 9 kilonewtons.

Maximum lap belt loads at each 'G'
(Load in kilonewtons)
(for all test conditions)



	9		16		20
Belt load	3.26		6.15		8.91

Designated 'G' level

— Belt load

Figure 5:33.2

Analysis of variance for lap belt force link

Table 5.33.3 outlines the findings from the analysis of variance. The loads generated in the lap belts did not appear to vary with initial belt tension.

Table 5.33.3

Source of variation	Significance (p=<)		Effect on force link
Brace position	0.001		Increased with braced
Leg position	0.05		Decrease with forward
'G' level	0.001		Increase with G
Brace and 'G'	0.01		Increase braced + G Increase unbraced + G
Leg and 'G'	0.01		Increase back + G Increase forward + G
Brace, leg and 'G'	0.05	9G	Increase unbraced + back Decrease braced + back
		16G	Increase unbraced + back Increase braced + back
		20G	Increase unbraced + back Increase braced + back

ii) Right knee shear

Because of technical difficulties only the results from the right knee sliding potentiometer (RL) were analysed. The means and standard deviations for each run type are recorded in table 5.33.1 in newtons (N). A graph of the maximum recorded knee shear load at each 'G' level is recorded in figure 5.33.4. Recordings from the potentiometer was seen to increase with the designated 'G' level. At 20G runs the load appears to increasing in perhaps an exponential manner.

Analysis of variance for right knee shear

The most important association for knee shear is the presence or not of flail. Table 5.33.5 illustrates residual associations which alter knee potentiometer readings if the

Maximum right knee shear at each 'G'
(Load in newtons)
(for all test conditions)

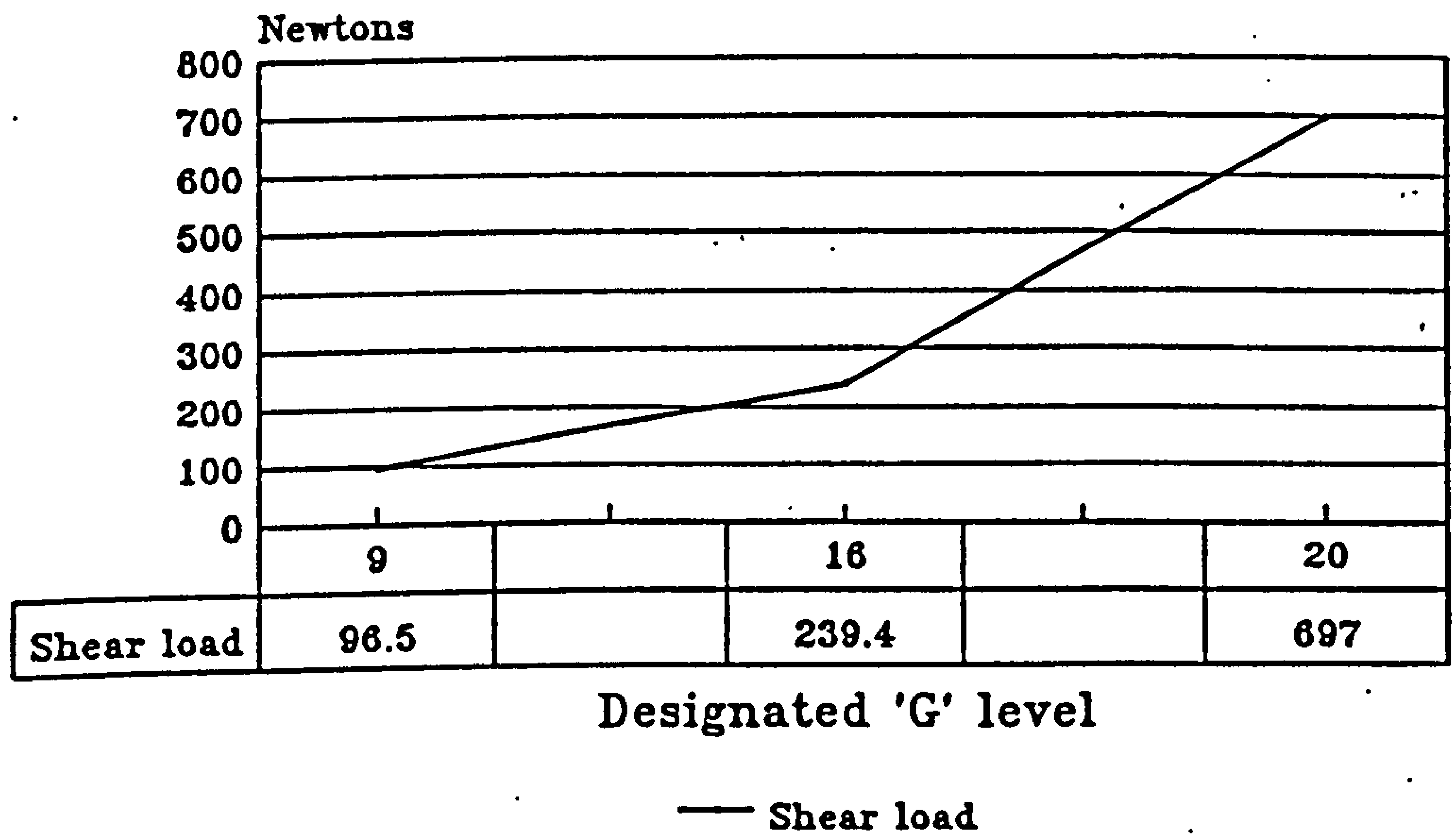


Figure 5.33.4

Table 5.33.5

Source of variation	Significance (p=<)		Effect on knee shear
Brace position	0.001		Increase with braced
Leg position	0.01		Increase with forward
Brace and 'G'	0.01		Increase braced + G
Leg and 'G'	0.05		Decrease back + G Increase forward + G
Brace, leg and 'G'	0.001	9G	Increase unbraced + back Increase braced + back
		16G	Increase unbraced + back Increase braced + back
		20G	Increase unbraced + back Increase braced + back
Leg and strap	0.05		Increase forward < tension Increase back > tension
Flail	0.001		Increase with flail

These effects are residual to the flail association.

uncontrolled variable of flail is removed from the analysis.

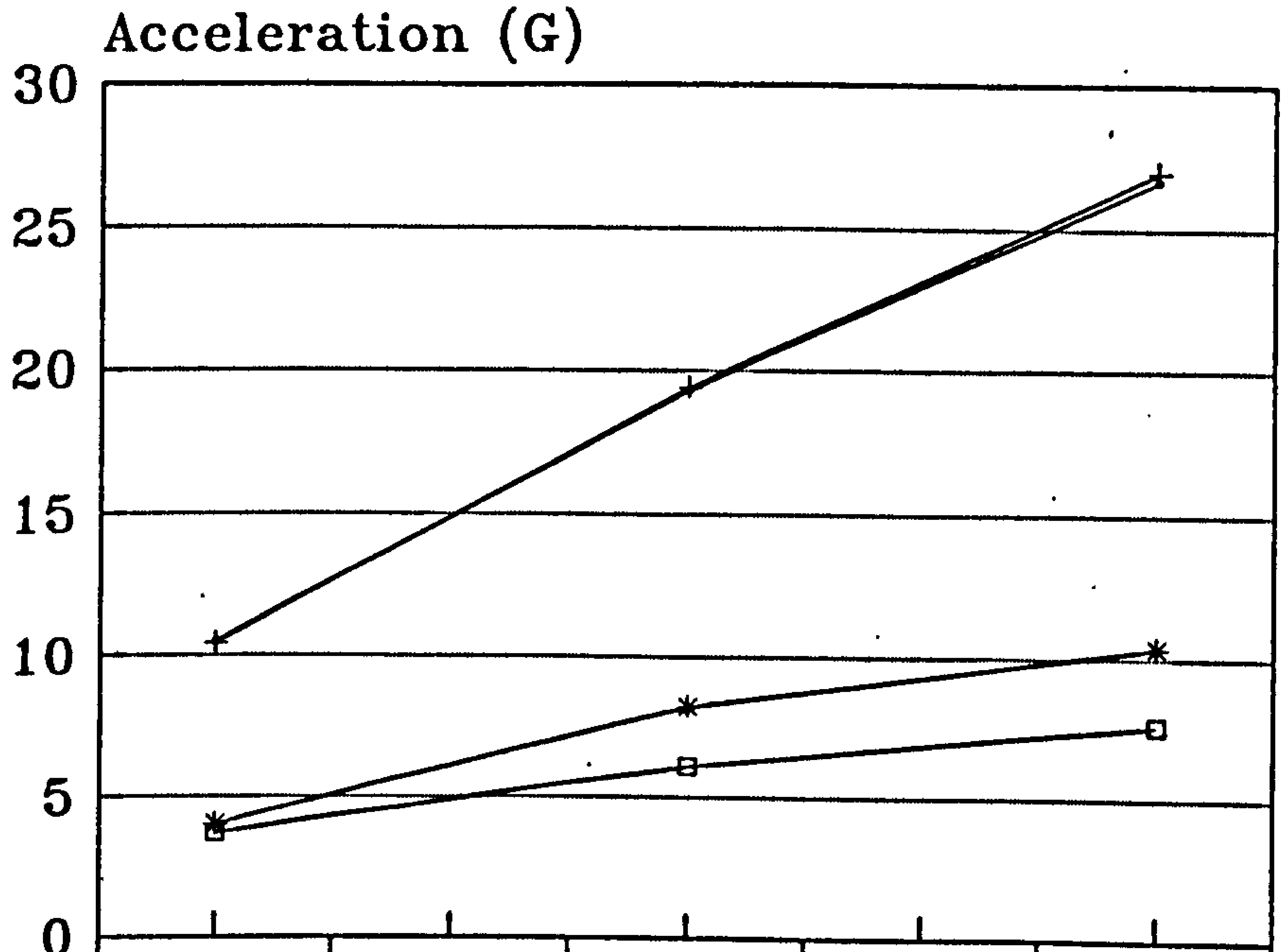
iii) Pelvic loads

The pelvic load cell measured pelvic horizontal (-Gx) acceleration and vertical acceleration (-Gz and +Gz). The vertical acceleration vector was shown to vary in different test situations with either a net footwards (-Gz) acceleration (PGzp) or a headwards (+Gz) acceleration (PGzn). For the analysis of the results each condition will be considered separately. However it must be remembered the vectors measured are relative to the initial placement of the load cells in the pelvis. On impact the pelvis as well as translating forward may rotate thus changing the orientation of the pelvic load cells. As a result of the rotation of the pelvis the Gx vector will come to lie in a more vertical plane and the Gz vector in a more horizontal plane. A more valid measure of the true level of pelvic acceleration may therefore be the resultant pelvic acceleration. The calculation of the resultant pelvic acceleration (PGres) has been described previously.

Table 5.33.1 demonstrates the means and standard deviations for all test conditions.

Figure 5.33.6 plots maximum pelvic accelerations at each designated 'G' level for pelvic Gx, pelvic Gres, pelvic +Gz

Maximum pelvic acelerations at each
(Acceleration in 'G')
(for all test conditions)



	9		16		20
Pelvic Gx	10.5		19.3		26.7
Pelvic resultant	10.44		19.4		27
Pelvic +Gz	4		8.2		10.35
Pelvic −Gz	3.7		6.1		7.6

Designated 'G' level

Pelvic Gx — Pelvic resultant
Pelvic +Gz — Pelvic −Gz

Figure 5.33.6

and pelvic -Gx. It can be seen that the maximum resultant pelvic accelerations recorded were similar to the maximum pelvic backwards acceleration (PGx) recorded. Both increase in a linear fashion with 'G' level. Pelvic backwards acceleration (PGx) and the resultant pelvic acceleration (PGres) were found to be greater than the designated 'G' level in all situations, with an increasing difference at higher 'G' levels. Pelvic loads can also be seen to be affected by flail behaviour. Both maximum pelvic headward acceleration (+Gz) and pelvic footward acceleration (-Gz) increase with G level, but are not dependent on the presence of flail behaviour but rather the positioning of the torso.

Analysis of variance for pelvic backwards acceleration

Pelvic acceleration in the horizontal (X) plane was found to vary with the following variables (table 5.33.7).

Table 5.33.7

Source of variation	Significance (p=<)	Effect on pelvic backwards acceleration
Brace position	0.001	Increase with unbraced
Leg position	0.05	Increase with forward
'G' level	0.001	Increase with G
Brace and leg	0.001	Increase unbraced + back
Brace and 'G'	0.001	Increase unbraced + G Increase braced + G
Flail	0.001	Increase with flail

Analysis of variance for pelvic headwards acceleration

Pelvic acceleration in the +Gz plane (PGzn) corresponds to a headward acceleration or eyeballs down. Table 5.33.8 illustrates the important associations of P +Gz with the following experimental parameters.

Table 5.33.8

Source of variation	Significance ($p=<$)	Effect on pelvic headwards acceleration
Brace position	0.001	Increase with unbraced
Belt tension	0.01	Increase with > tension
'G' level	0.001	Increase with G
Brace and 'G'	0.001	Increase unbraced + G Increase brace + G

Analysis of variance for pelvic footwards acceleration

Pelvic footwards acceleration (PGzp) was found to vary with the following parameters as indicated in table 5.33.9 below.

Table 5.33.9

Source of variation	Significance ($p=<$)	Effect on pelvic footwards acceleration
Brace position	0.001	Increase with braced
'G' level	0.001	Increase with G
Brace and leg	0.01	Increase unbraced forward Increase braced + back
Brace, leg and 'G'	0.001 9G	Decrease unbraced forward Decrease braced + back

16G	Increase unbraced forward	
	Increase braced + back	
20G	Increase unbraced forward	
	Increase braced + back	

Analysis of variance for pelvic resultant acceleration

Pelvic resultant acceleration is a measure of the maximum acceleration experienced by the pelvic load cell. Table 5.33.10 illustrates the important associations seen with the experimental parameters investigated.

Table 5.33.10

Source of variation	Significance (p=<)	Effect on pelvic resultant acceleration
Brace position	0.001	Increase with unbraced
Leg position	0.05	Increase back
'G' level	0.001	Increase with G
Brace and leg	0.01	Increase unbraced + back
Brace and 'G'	0.001	Increase unbraced + G Increase braced + G
Flail	0.001	Increase with flail

5.4 Summary of Results

Flailing of the lower limbs is associated with both leg position and with bracing ($p<0.001$ in both cases). In no trials when the legs were placed in a 'back' position did the legs flail. When the dummy was placed in a braced position, only at designated G levels of 9G and 16G was no flailing seen, although on five occasions at the highest

'G' level flailing of the lower limbs was seen.

The designated 'G' level effected almost all measurements, such that with increasing G the measurement increased. The notable exception being that of ankle vertical displacement that decreased with G in the presence of flailing.

Flailing of the lower limbs has a significant effect ($p<0.001$) on horizontal ankle displacement (A dX), ankle vertical displacement (A dZ), knee vertical displacement (K dZ), right knee shear, pelvic backwards acceleration (PGx), and the resultant pelvic acceleration (PGres). Excluding flail behaviour as an uncontrolled variable in further analysis the following general effects are observed.

Horizontal displacement of the pelvis is greater in the unbraced population ($p<0.001$), higher if the belt has a low initial tension ($p<0.001$) and greater in positions with the legs forward in the unbraced position ($p<0.001$). Vertical displacement of the pelvis is higher in the braced position ($p<0.001$).

Contact of the knee with the back of the seat in front did not occur in any of the tests until an extra factor of a lap belt tension of 0lbs was investigated in the unbraced , feet forward situation. It then only occurred at the 20G level. Knee displacement in the horizontal plane is greatest in situations when the lap belt initial tension is low ($p<0.001$), and is of smaller magnitude than pelvic horizontal displacement (P dX) and ankle horizontal displacement (A dX) for all parameters investigated.

Vertical knee displacement varies with many factors in a way similar to that of right knee shear, but there is an obvious effect of flail ($p<0.001$) in both cases.

Measured pelvic horizontal acceleration (PGx) and resultant pelvic acceleration (PGres) are greater for unbraced ($p<0.001$) than the braced positions. There is also a small effect of increased accelerations in the unbraced position if the legs are placed in a back position ($p<0.01$). Pelvic headward acceleration increases in the unbraced conditions ($p<0.001$), but an increased pelvic footwards acceleration was seen in the braced position ($p<0.001$).

Lap belt force link recordings increase with a braced position ($p<0.001$) and are decreased in position with the legs forward ($p<0.001$), in the absence of flailing.

5.5 Analysis and Discussion of the Results

5.51 Motion of anthropomorphic dummy

From the video recorded evidence the flail behaviour of the lower limbs could be seen to be modified by the position in which the ATD is placed. If the feet were placed slightly behind the vertical axis of the knee (12 degrees) no flailing was seen in all test situations at all 'G' levels investigated. Further in no trials at 9 and 16G (designated level), where the dummy was placed in the braced position did the lower limbs flail. However this effect is inconsistent at higher levels of 'G'.

Coltman in 1982 (cited Chandler 1990), investigating vertical +Gz energy absorbing seats for helicopters, concluded that the placement of the feet and lower limbs can significantly influence seat and occupant response in a dynamic test. It can also be concluded positioning of an anthropomorphic test device's lower limbs can significantly affect flailing of the lower limbs under the seat in front if a horizontal -Gx vector is considered. If this affect can be demonstrated at other impact vectors (perhaps those recommended by the FAA (1989)) it has obvious implications in reducing injuries seen as a consequence of flailing behaviour seen as a result of horizontal decelerations.

5.52 Selspot displacement data

Flail and non flail behaviour have been identified from the ankle displacement recordings, as not occurring when the feet are placed behind the knees. When the feet are placed in front of the knee joint no flailing was seen at the 9 or

16 G, however at 20G flailing was seen in some tests.

Contact of the knee with the knee panel of the seat in front was not identified in any of the test conditions at a 32 inch seat pitch. However when an additional parameter of a lap belt tension of 0lbs was investigated in the test condition of unbraced, legs forward (runs 1123, 2123 and 3123), it was found that slight knee contact occurred in only the 20G runs. This suggests that significant axial loading of the femur may not occur as a result of knee impact with the seat in front.

Horizontal ankle motion appears to reach a maximum value at 20G, whereas ankle vertical displacement decreases with increasing G levels. Maximum knee displacement in the horizontal plane increases in a linear fashion with G, but vertical knee displacement appears to reach a limit. For pelvic displacement, again a maximum excursion seems to be reached. Vertical pelvic displacement appears to increase in a linear fashion with G.

Where displacements tend towards a maximum it suggest that further motion has been limited or restrained. Figure 5.52.1 illustrates possible factors that limit motion. At higher designated 'G' levels pelvic motion is limited by the lap belt as the natural distensibility is taken up. Knee vertical displacement is limited by the anterior

Limits of pelvic, knee, and ankle movements

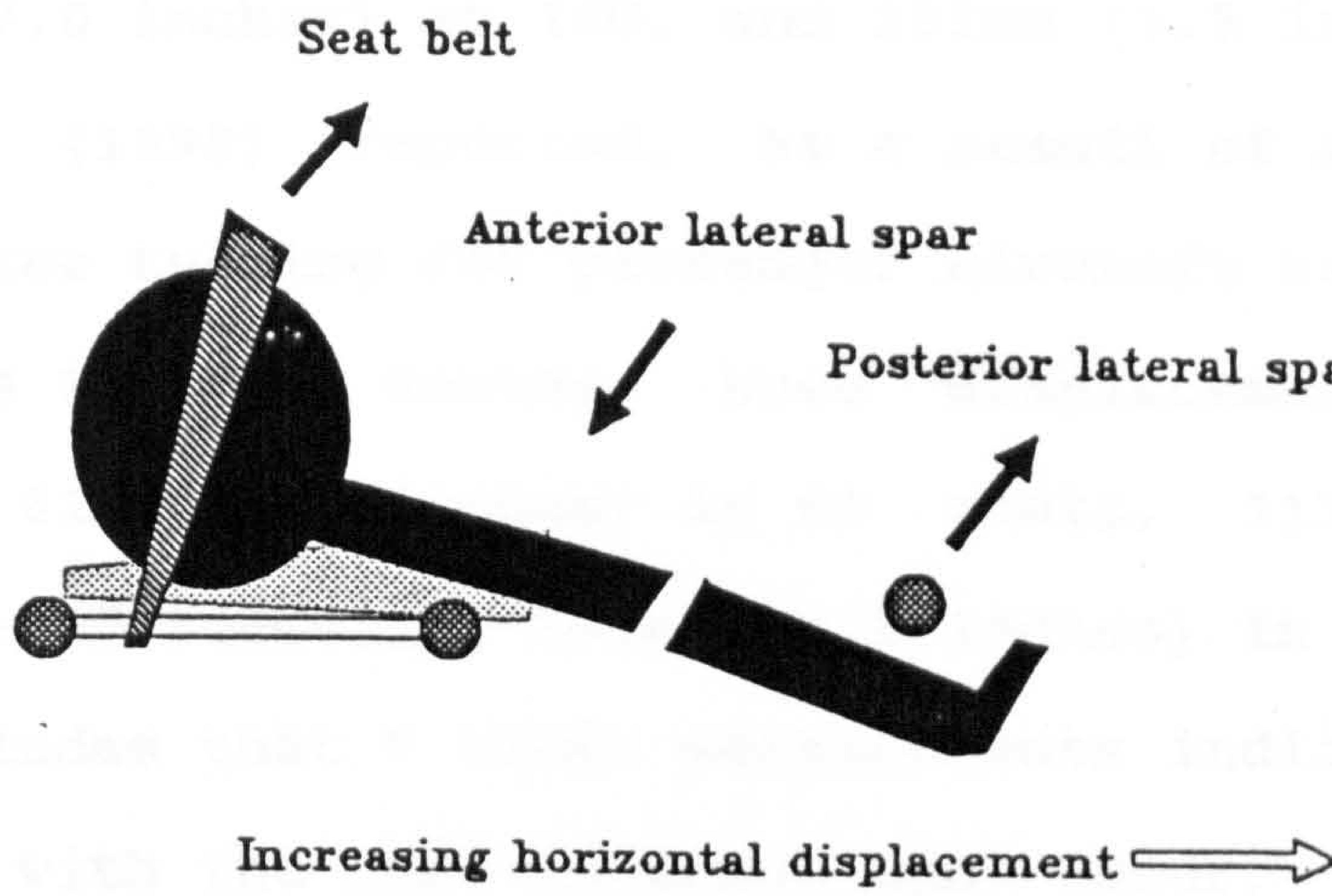


Figure 5.52.1

lateral spar of the seat. Ankle horizontal displacement is limited by contact of the shin with the posterior lateral spar of the seat in front and by the lap belt restraining the occupant. With increasing knee and pelvic horizontal displacement the contact point of the shin with the seat in front becomes more proximal on the shin (nearer the knee). This has the effect of limiting the amount of vertical excursion of the ankle. As a consequence of this effect vertical knee displacement can also be seen to be increased. This effect is only apparent in situations of flail. If motion is limited by contact with a fixed structure a load must therefore be applied.

Maximum horizontal knee excursions recorded at each G level (runs 1123, 2123 and 3123) were 162 mm (6.3 inches) at 9G, 178mm (7.0 inches) at 16G, and 191mm (7.5 inches) at 20G. Chandler (1990) reported, as a result of seat and restraint system testing for passenger aircraft using two rows of seats in a -Gx vector, knee displacements were found to be 81mm (3.2 inches) in 6G tests, 117mm (4.6 inches) in the 9G teats and 132mm (5.2 inches) in the 12 G tests. He concludes that " these measurements indicate that knee contact with the seat in front can occur, and may cause injuries as well as introduce unexpected loads on the seat". The results of this study indicate that at a 32 inch (81.3 cm) seat pitch, contact of the knee with the seat in front was not significant even though our knee displacements were greater than those measured by Chandler. Significant axial loading of the femur through the patella region does therefore not occur in these experimental conditions.

Pelvic displacement in the horizontal plane was seen to increase in an unbraced position. This may reflect the greater inertia created on the pelvis by the torso and head 'jack knifing' around the lap belt, whereas in the braced position the head and upper limbs are braced against the seat in front. This effect was increased with a lower initial tension in the lap belt. It was apparent that the pelvis 'rotated' on impact and is indicated by greater pelvic horizontal displacement measurements being recorded than that of horizontal knee displacement.

Vertical displacement of the pelvis increased with G and also with the adoption of a brace position. This may reflect less pelvic translation at impact but more motion in line with the lap belt and rotation around the lap belt as a result of the torso striking the thighs, fixed because flailing of the lower legs has not occurred, and because further forward motion of the torso was limited by contact with the seat in front. Thus in the braced position greater pelvic vertical displacement was seen whereas in the unbraced position greater horizontal displacement was seen.

Horizontal knee displacement was seen to increase with G level and increase with a slack belt (tension 20lbs). Forward translation of the femur resulted in the more proximal portions of the thigh lying over the anterior lateral spar of the seat.

The main effect on vertical knee displacement was in the flail situation. Flailing results in vertical knee motion of up to 126 ± 4.4mm, however in the non flail situations the maximum displacement is 45 ± 26mm at the highest designated G level. (These are displacements measured from a LED placed 22cm from the knee). Thus in situations where the lower limbs do not flail the thigh may cause compression of the seat squab but insufficient to cause significant loading across the anterior lateral spar of the seat. In this situation vertical knee displacement is prevented, the lower leg acting as a strut to prevent downward motion of the knee. In the flail situations large vertical displacements of the knee were seen. With increasing displacement one would expect significant loading of the femur over the anterior spar.

In the absence of flail behaviour vertical knee displacement is greatest in positions with the legs forward. This is facilitated partly by an increased horizontal translation of the foot on the floor of a degree greater than that of horizontal knee displacement.

Ankle displacement in both planes is found to vary with brace position, leg position, designated 'G' level and with flail behaviour. Further ankle displacement in the vertical plane was seen to be greater if the legs were forward and the lap belt tight. In this situation less pelvic horizontal displacement resulted in the shin striking the seat ahead more distally, thereby allowing greater vertical displacement of the ankle.

5.53 Datalab recorded data

The maximum loads recorded by the lap belt force link were in the order of 9 kN. Loads were found to increase in all experimental conditions investigated with G. Increased lap belt loads were associated with the adoption of a braced position and positioning of the feet behind the vertical axis of the knee joint. Rotation and translation of the pelvis in line with the lap belt, augmented by a fixed lower limb (acting as a strut) has probably resulted in the increased loads seen. Decreased lap belt loads were recorded when the lower limbs were placed in a forward position.

The lap belt passes over the greater trochanters of the femur and below the anterior superior iliac spine of the pelvis. King (1985) in his review of the 'Pelvis' identified acetabular fractures, pubic rami fractures, sacroiliac joint injuries, iliac wing fractures and proximal femur fractures as occurring as a result of lateral loading with an impactor to the greater trochanter of the femur, at a range of 4.4 to 12.9 kN. These injuries have been described as result of lateral impacts in automobiles and falls. The lap belt tensions generated in the tests suggest that sufficient loads may be transferred from the lap belt to the pelvis to cause injury.

The right knee shear potentiometer measures tibial displacement on the femur. In the experimental protocol impact of the knee against the seat in front was expected. However it was found that only when an additional parameter of a belt tension of 0lbs was investigated, at 20G, was knee contact observed and this was only sufficient to cause minimal indentation in the plasticine witness indicator placed on the knee panel of the seat in front.

The results indicated that the most important factor affecting knee potentiometer readings was the affect of flail. When flailing occurred maximum knee potentiometer readings were seen. For all test conditions the greatest mean loads recorded were 700 ± 390 N. In those conditions that resulted in flailing at the 20G level, examination of the video recordings indicate hyper-extension of the knee assemble of the ATD. Indeed this effect was so severe as to shear the knee assemble bolts in the ATD placed in the forward seats.

Noyes and Grood (1976) demonstrated that an isolated anterior cruciate specimen failed in a range of 622N to 1170N. The loads generated in this experiment may have been sufficient to cause injury in particular to the posterior cruciate ligament, which is in tension with extension of the knee, and also in posterior displacement of the tibia in relation to the femoral chondyles as seen in bolster impacts to the knee.

In the presence of flailing, knee potentiometer recordings were greatest with low lap belt tensions. In these conditions a greater horizontal knee displacement was evident, resulting in a more proximal strike of the ATD's lower leg (shin) against the posterior longitudinal spar of the seat in front. A greater moment may therefore be created by the ankle flailing under the seat in front. (See figure 5.53.1). There was also the additional effect of torso rotation, which was seen to increase vertical displacement of the knee, producing hyper-extension.

Factors effecting knee shear recordings

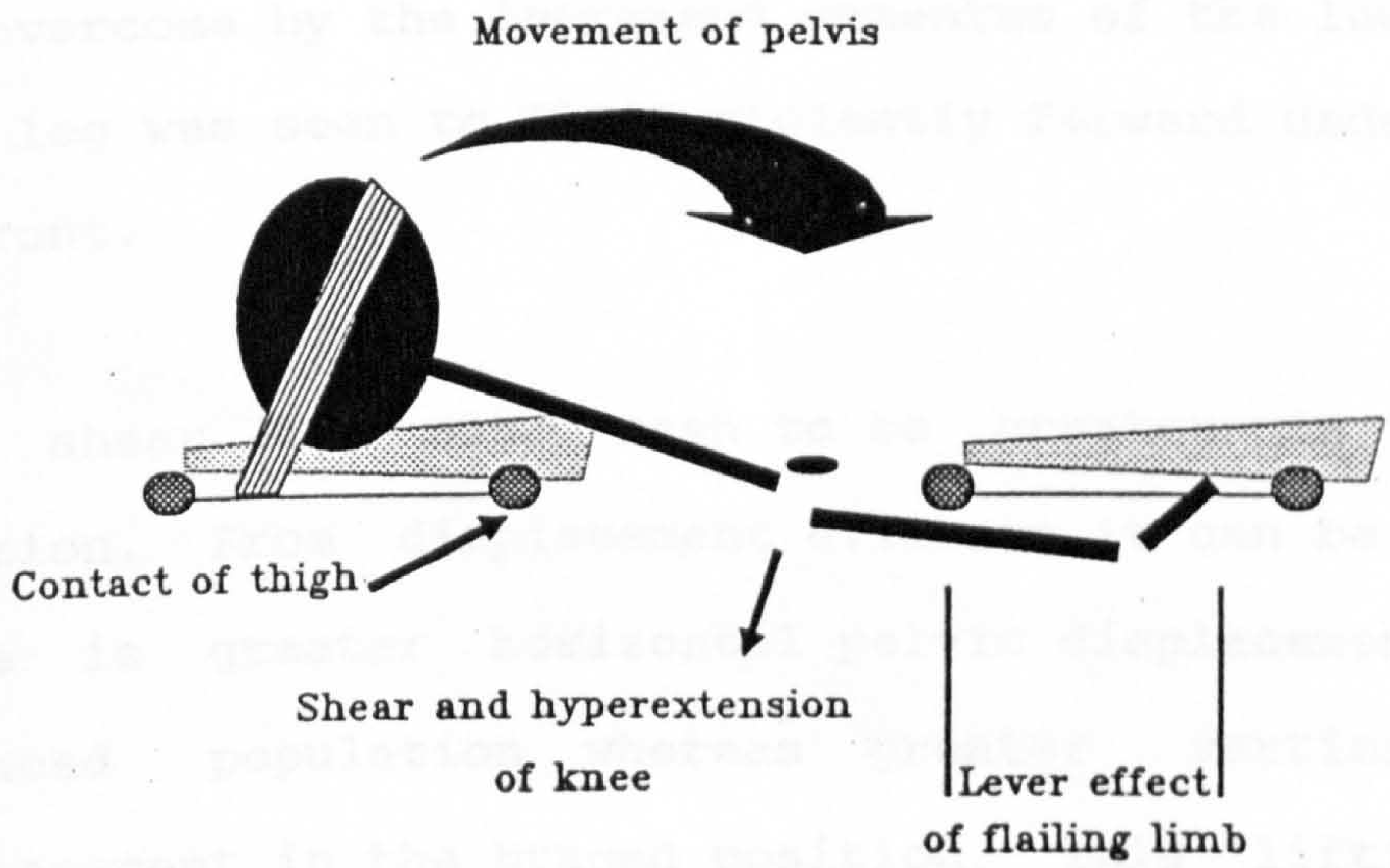


Figure 5.53.1

In those situations where flail did not occur knee shear was seen to be greatest in the leg forward tests. On impact the ATD's torso was seen to rotate round the lap belt. In the absence of flailing the torso came into contact with the thighs which resulted in compression loading of the lower leg, which was planted on the floor. The effect of this, with the leg in a forward position, was an attempt to straighten (extend) the knee. Flailing was prevented because of friction between the shoe and the floor, and in this way shear was induced in the knee. This effect was confirmed by video analysis. The foot was seen to 'sutter' along the floor, with some of the test runs demonstrating a small flip at the end of the excursion. Displacement data confirms these observations. With high G levels this effect was overcome by the increased momentum of the lower leg and the leg was seen to flail violently forward under the seat in front.

Knee shear was also seen to be greater in the brace position. From displacement evidence it can be seen that there is greater horizontal pelvic displacement in the unbraced population whereas greater vertical pelvic displacement in the braced position. This 'lifting affect' with bracing, would tend to vertically displace the knee (downwards) as the femoral head is lifted by the pelvis, a lever effect attempting to straighten the knee, being prevented by a planted foot. The anterior lateral spar acts as a fulcrum for this motion to occur. (see figure 5.53.2).

In those situations where there was decreased horizontal displacement of the pelvis, the anterior lateral spar of the seat can be seen to be more distally located in relation to the thigh. With vertical displacement of the pelvis (lift) a smaller distal lever is present and therefore less vertical knee displacement would be possible (figure 5.53.2). In this situation knee shear recordings were seen to be small.

The anterior lateral spar as a fulcrum affecting vertical knee displacement

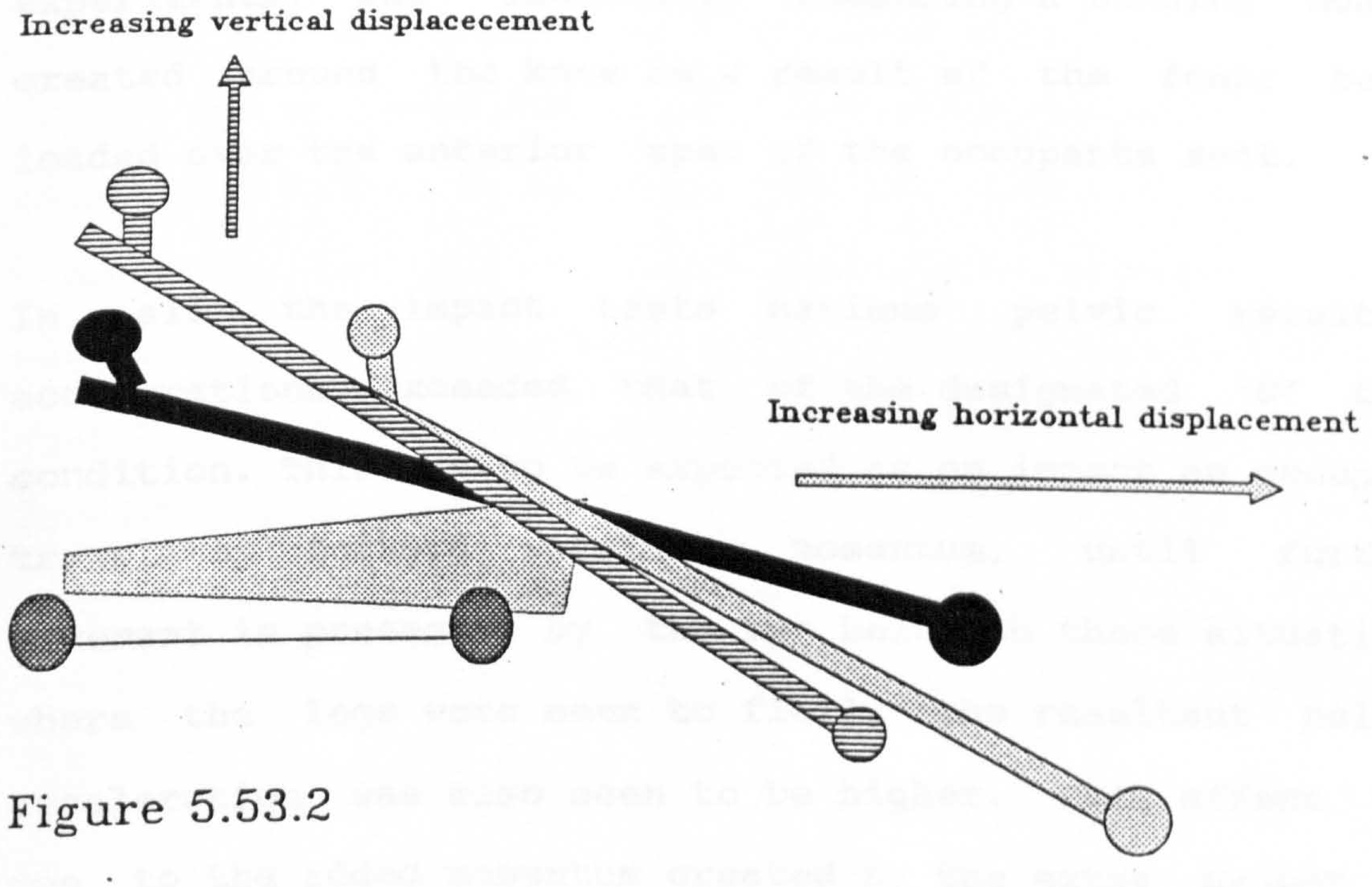


Figure 5.53.2

Thus vertical knee displacement was found to be an important factor in producing elevated knee shear recordings, the anterior lateral spar of the seat acting as a fulcrum (figure 5.53.2). Test conditions which result in an increase in knee vertical displacement will thus tend to result in greater knee shear readings. The most important effect is that of flailing of the lower leg. Further it can be seen that with increasing vertical knee displacement the thigh can be seen to be loaded over the anterior lateral spar.

Knee shear potentiometer recordings were thus not occurring as a result of posterior tibial displacement on the femur, caused as a result impacts to the region of the knee. It appears however that the knee potentiometer, in the experiments, was indirectly measuring a bending moment created around the knee as a result of the femur being loaded over the anterior spar of the occupants seat.

In all the impact tests maximum pelvic resultant accelerations exceeded that of the designated 'G' test condition. This was to be expected as on impact an occupant translates forward attaining momentum, until further movement is prevented by the lap belt. In those situations where the legs were seen to flail, the resultant pelvic acceleration was also seen to be higher. This affect was due to the added momentum created by the extra weight of flailing lower limbs. There was also a residual effect, in the absence of flailing, indicating that pelvic loads were slightly higher if the legs are placed in the 'back' position. In the forward position because the legs act as a 'strut', they act to oppose displacement of the pelvis. However this effect is small.

Pelvic horizontal accelerations mirror many of the findings of the resultant pelvic accelerations. Flailing behaviour is the most significant factor affecting accelerations. Interestingly the initial tension of the lap belt does not seem to affect these parameters.

Headward pelvic acceleration was however affected by the initial tension of the lap belt, such that with greater tension (40lbs) greater accelerations were recorded. This effect was not seen for pelvic footwards acceleration. The main conclusion to be drawn from analysis of these two components is that unbraced conditions produce maximal headward accelerations (the pelvis 'sinks' into the seat), whereas in the braced position the pelvis produces maximal accelerations in the footwards direction (the pelvis 'rises' in the seat). These effects relate to the relative motions of the torso in differing brace positions, restrained by a lap belt.

If one proposes that the pelvis produces a maximal footwards acceleration in the braced position one would expect greater lap belt forces in this position, and pelvic vertical displacement. This was indeed the case. These factors have been found to affect vertical knee displacement and therefore indirectly loading of the thigh over the anterior lateral spar of the occupants seat.

5.6 Conclusions

Impact biomechanics, for the pelvis and lower limbs, have been investigated using a twin seat row configuration and an instrumented Hybrid III anthropomorphic test device. The experimental protocol investigated the effects of the position adopted at the time of impact, and the effect of differing lap belt tensions. Three designated 'G' levels were investigated in a -Gx horizontal vector at 9G, 16G and 20G.

Flailing of the lower limbs under the seat in front has been shown not to apply in all test situations. Positioning the feet so that they lie behind a vertical line drawn through the knee, prevents flailing of the lower leg.

Flailing occurs in those situations where the feet are placed in front of the knee joint. In flail situations the lower leg can be seen to strike the posterior longitudinal spar of the seat in front, and the foot the sub seat assembly. It has also been demonstrated that contact of the dummy's knee with the seat in front does not occur at a 32 inch seat pitch, indicating axial loading of the femur, as a result of such an impact, has not occurred. An instrument panel type mechanism of femoral and hip injuries was not apparent in the experiments.

The experiments have indicated that the femur was being loaded over the anterior lateral spar of the seat. The motion of the lower limbs and pelvis was seen to be limited by three fixed points, the lap belt, the anterior lateral spar of the seat and the posterior lateral spar of the seat in front (see figure 5.52.1 and 5.53.1). Thus it can be seen the thigh was subjected to three point loading.

The knee shear potentiometer recordings, in the absence of bolster impacts to the knee, gave an indirect measure of the bending moment created around the knee. The loads generated across the knee potentiometer would be sufficient to cause ligament injury around the knee.

From the results it can be seen that the thigh is loaded over the anterior lateral spar of the occupants seat creating a bending moment in the femur. This bending of the femur was seen to be greatest in situations of flail and also greatest in a braced position with the legs placed in a forward position. In addition factors that increase the amount of forward motion of the knee (low lap belt tension) tend to increase the amount of knee shear seen and therefore femoral bending.

Increased vertical knee displacement has been found to increase the knee potentiometer readings. Methods that reduce vertical displacement of the knee will decrease knee shear recordings and therefore the amount of loading of the femur over the anterior lateral spar of the seat.

Tensions generated in the lap belt may be sufficient to cause injury by compression of the pelvis and loading of the greater trochanters of the femur.

Chapter 6

Interpretation of Results and Conclusions from both the Clinical Reviews and the Impact Testing

6.1 Introduction

In this chapter the mechanisms for the pelvic and lower limb injuries sustained by occupants, seated in the intact mid section of the aircraft, are proposed and explored. Evidence from the clinical review of the injuries sustained has been combined with the findings from experimental impact testing using a decelerator track. The mechanisms of injury in the severely damaged sections of the aircraft have not been considered further because the structural failure of the airframe and seating made detailed analysis impossible.

Anthropomorphic test devices (ATD'S) have been used extensively in the automobile industry to assess impact biomechanics of occupants involved in automobile accidents particularly for frontal impacts. The biofidelity or likeness to real life of anthropomorphic test devices has been criticised in the past and they have been referred to as 'only sophisticated dolls'. In addition their response to vertical loading, which may occur as a result of aircraft accidents, is criticised. Since all commercially available ATD's were developed for use in the atomobile industry their response to impact has been optimised for the horizontal Gx axis impacts. Fidelity is limited in lateral Gy directions and the response of the dummies are generally stiffer than that of humans in the vertical Gz vector. Appendix 3 lists many of the important anthropometric dimensions of the ATD and demonstrates their similarity to those of the occupants seated in the middle section of the aircraft when it crashed. In particular there was a good correlation between the ATD (50% Hybrid III) and the occupants for the buttock knee length and seating knee height.

In the impact tests, using a linear decelerator track, designated accelerations in a horizontal (-Gx) attitude only were used. The tests were thus only able to simulate the acceleration change in this vector.

In the previous chapters the general mechanisms of the pelvic and lower limb injuries have been suggested. The detailed mechanism of injury is now explored for the pelvis; knee-femur-pelvis complex; and the lower leg for occupants seated in the intact central section of the aircraft. It must be remembered that the mechanisms of injury causation described cannot be treated as separate entities but considered in part as interacting with other factors. The effects of loads applied in the Gx and/or Gz vectors on injury causation cannot necessarily be considered in isolation, but may have additional effects on the mechanisms described.

6.2 Pelvis

Pelvic injuries are caused as a result of external forces that are applied either directly to the bony structure of the pelvis or are transmitted through the femur. This section will first review those injuries resulting from forces applied directly to the bony structure.

The external forces acting on the pelvis are those transmitted through the seat structure and those acting through the seat belt as well as inertial forces transmitted through the lumbar spine as a result of upper body flailing. A review of the literature has revealed that sacro-iliac diastasis and pubic rami fractures or diastasis are commonly seen following aircraft accidents in which there are high vertical loads (Mason 1962, Gillies 1965, Hill 1984). Such injuries are caused as a result of loading of the pelvis and spine through the seat, a vertical shearing force (Tile 1988) (figure 2.33.1). It has been estimated, the occupants of the middle section, experienced a vertical component of acceleration of +20 to 25 Gz.

However additional loads may also be applied to the pelvic ring. Tile (1988) has stated that internal rotation (lateral compression) of the pelvic ring, from a direct blow over the lateral aspect of the pelvis, or indirect force through the femoral head may result in pelvic ring fractures. Lateral impact to the pelvis from automobile impact trauma is known to result in pelvic rami fractures, iliac crest fractures and central dislocations of the hip (intrapelvic fractures) (Gratton and Hobbs 1969, Epstein 1973, Walz 1984, Dejeammes 1984, King 1985, States 1986 McCoy et al 1989). Cadaveric studies fo dynamic lateral loading of the greater trochanter have also demonstrated such fractures at loads of between 4.4 to 12.9kN.

The lateral component (±Gy) of the acceleration vector of the Kegworth aircrash was minor and it is unlikely that a direct impact over the lateral aspect of the pelvis was significant. However impact testing has demonstrated loads generated within the lap belt are as high as 9kN, and severe bruising of the pelvis as a result of the lap belts did occur in this accident (Rowles et al 1990). The loads generated in the lap belts were sufficient to cause direct injury to the bony pelvic ring by either an internal rotation (lateral compression), or external rotation (open book) (figure 2.33.1) mechanism. Internal or external rotation of the pelvic wings may depend on individual variation such as sex and therefore pelvic shape. Diastasis of the symphisis pubis indicates an external rotation mechanism, whereas pubic rami fractures with sacroiliac ligamentous damage indicates internal rotation (Tile 1988). Indeed all of the iliac crest fractures must have been due to the loads transmitted from the lap belt.

The geometry of the lap belt in the seats under study was such as to cause it to pass over the greater trochanter of the femur. Loads of up to 9 kN were generated within the lap belts during experimental impact testing. These loads are sufficient to cause central hip dislocations or intrapelvic fractures as a result of loading over the greater trochanter of the femur.

6.3 Knee-femur-pelvis complex

Research work carried out in the automobile industry has shown that axial loading of the femur is known to cause injuries to the knee, femur and pelvis (Ritchey at al 1958, Grattan 1969, Viano et al 1980, Viano 1980, Viano and Stalnaker 1980, Chapon 1983, Cheng et al 1984, King 1985 a+b, Nyquist and King 1985, Viano and Levine 1986, States 1986, McCoy et al 1989). As a result injury criteria, or loads at which injury can be expected to occur, have been identified for the pelvis and lower limbs. Such injury mechanisms have also found wide acceptance in the aviation industry and as a result injury criteria have been set for occupant protection safety systems design in aircraft and are detailed in FAA Advisory circular AC No: 21-22 (Pontecorvo 1985). Many of the recommendations for occupant safety system design and load levels likely to result in injury have thus resulted from research in the automobile industry.

It is also apparent that many of the studies with cadavers have failed to reproduce the type of femoral fractures most commonly seen in impact accidents, but result in a preponderance of fractures of the distal femur (Nyquist and King 1985). Femoral fractures have been described as a result of the knees sliding under the car dash board (Ritchey at al 1958, Nyquist and King 1985, States 1986).

In the sections of the aircraft where seating failures occurred, indentations in the backs of the seat, and damage to the knee panels was evident. Similar indentations were seen in the back of seats in the centre section of the aircraft although less severe damage was apparent to the knee panels. Where seating concertinaed the damage to the seat backs and knee panels was likely to have been caused by the impact of the knees of the occupant behind.

However, clinical review of the occupants seated in the mid section of the aircraft, suggest no association of soft tissue witness marks to the knee nor to injuries associated with axial loading of the femur eg. femoral fracture. In addition it is apparent, in those individuals with the greatest buttock to knee lengths, soft tissue injury around the knee did not occur. Impact testing indicated that significant impact of the knee with the seat in front did not occur in any test conditions up to 20 G. It can therefore be postulated that axial loading of the femur has not contributed significantly to those injuries usually associated with the knee-femur-pelvis mechanism (instrument panel syndrome). High speed video of the deceleration track tests has demonstrated that the indentations seen in the seat backs occurred as a result of the head and upper limbs impacting with the back of the seat in front.

If axial loading of the femur is not the primary mechanism of the injuries seen as a result of axial loading what was the mechanism ?

The clinical review of the injuries revealed that the majority of femoral fractures occurred in the proximal portions of the femur. In addition there was an increased incidence of femoral fractures in the centre seat of a row of three seats, where the anterior lateral seat spar was found on mechanical testing to be more rigid as a result of being supported at either end. In the outer seats the anterior lateral spar was frequently bent, on the unsupported side (figure 4.3.2.1) as a result of the forces generated in the crash. It was also noted, although not statistically significant, that there was an increased incidence of femoral fractures in those occupants who adopted a brace position and with those who had longer than average femurs.

The impact testing experiments revealed that the lower limb and pelvis was loaded by contact with three fixed points i) the lap strap ii) the posterior longitudinal spar of the seat in front and iii) the front spar of the occupants seat. Impact testing also demonstrated that hyper-extension of the knee occurred when the legs flailed. The important primary mechanism of femoral fractures was therefore most likely to be one of three point bending of the femur, rather than axial loading. The bending loads transmitted to the knee-femur-pelvis complex occurred as a result of the lap belt restraint; flailing of the lower legs under the seat in front and contact with the posterior spar of the seat in front; and the anterior spar of the same seat. The anterior spar of the seat also acted as a fulcrum for the proposed mechanism of three point bending of the femur. (figure 5.52.1 and 5.53.1).

The forces applied to the femur over the anterior spar, were demonstrated by impact testing to be dependent on the amount of vertical displacement of the knee and these forces were indirectly indicated by the amount of knee shear recorded by the knee potentiometer on the ATD. These effects were increased by the adoption of a brace position, and placement of the dummy's feet in front of the knee joint.

Those individuals with longer than average femurs were noted to have sustained more femoral fractures. Three point bending as a mechanism of fracture of the femur suggests that with a longer lever (femur) increased bending moments would occur.

Impact testing also demonstrated considerable forward translation of the ATD so that the more proximal part of the regions of thigh lay over the anterior spar. Again the clinical findings illustrated that fractures of the femur occurred almost universally in the proximal femur.

A number of occupants seated in the centre of the aircraft were found to have traumatic knee effusions and one individual suffered a ruptured posterior cruciate injury. Posterior cruciate injuries to the knee are described as a result of loading of the proximal tibia through the lower dash panel of automobiles (Viano et al 1978, States 1986), consequently the tibia is displaced posteriorly beneath the femoral condyles. All non boney knee injuries in occupants seated in the centre section of the aircraft were sustained in the absence of a soft tissue witness of impact to the region of the knee, although most demonstrate bruising to the region of the ipsilateral shin. Impact testing demonstrated hyper-extension of the knee with flailing. It is therefore most likely that injuries to the knee have resulted from forced extension of the knee following lower leg flailing. Forces recorded in the knee potentiometer during the impact testing (of the order of 1 kN) are recognised as being sufficient to cause rupture of the posterior cruciate ligament.

This research work has therefore demonstrated that even in the absence of a vertical component of acceleration in the impact tests carried out, mechanisms of injury can be identified that are corroborated by the clinical observations. However additional possible effects of loading in the Gz plane were unable to be simulated with impact testing.

Axial loading of the femur is recognised as being an important mechanism in producing posterior hip dislocation and acetabular fractures, as seen in automobile accidents.

If axial loading of the femur did not occur, was another mechanism responsible for the posterior hip dislocations and the acetabular fractures?

Impact testing has demonstrated that pelvic horizontal displacement was greater than femur horizontal displacement. This was as a result of rotation of the pelvis and torso around the lap belt. Clinical photography of the injured survivors has identified bruising to the proximal outer thighs (Rowles et al 1990). This may represent injury caused by the lap belt as a result of rotation of the pelvis. The flexion and rotation of the pelvis and torso, occurs around the femoral head and hip joint. This motion has the effect of pulling the torso forward on a femur fixed by the lap belt, thus causing the acetabulum to dislocate by forward motion of the pelvis on the 'restrained' femur. This mechanism may also potentiate the effect of vertical shear loading of the pelvis and pelvic ring injuries. The forces generated may also lead to posterior column fractures of the acetabulum, and posterior hip dislocation This biomechanical mechanism has not been previously described nor appreciated. In addition, if the thigh was in the abducted position, thus providing increased cover of the femoral head, shear fractures within the femoral neck or in the intertrochanteric region of the femur would be more likely. The mechanism of force application can be likened to that of a cantilever (Cockran 1982). Such fracture patterns as those discussed were seen in 3 of the occupants seated in the mid section of the aircraft.

6.4 The lower leg

Flailing of the lower limbs in impact aircraft accidents is important in the causation of injuries to the shin, ankle and foot (Swearingen et al 1961, Mason 1962, 1973, Gilles 1965, Stevens 1970, Horne and Mowbray 1980, Hill 1984).

Indeed most the injuries to the lower leg are due to this mechanism.

On impact the legs flail under the seat in front with the shin striking the posterior longitudinal spar while the feet and ankles strike the sub seat assemblies of the seat in front. As a result of the contact of the shin with the seat in front, fractures of the tibia and fibula frequently occur. The deformation suffered by the feet under the seat in front will result in hyper-extension of the ankle joint and fore-foot, resulting in fractures and dislocations of the fore-foot and ankle. Hyper-extension of the foot is also recognised as a cause of fractures of the talus (Hawkins 1970).

A careful clinical review of the injuries sustained by the occupants seated in the mid section of the aircraft revealed that some of the occupants had no soft tissue injuries to their lower legs suggesting that significant impact had not occurred. Impact testing on the decelerator track demonstrated that flailing of the lower limbs could be modified by the position of the limbs at the time of impact.

Tibial plateau fractures, tibia and ankle injuries are described in occupants of automobile accidents as being a result of the rearward movement of the toepan of the car (axial loading) coupled with torsion and/or bending moment (Nyquist and King 1985, States 1986). Studies have demonstrated that in the automobile industry flailing of the lower leg is not an important injury mechanism.

In the absence of flailing axial loading of the lower leg can occur with forces transmitted through the foot firmly placed on the floor. This mechanism may be applicable to occupants who sustained fractures of the os calcis and tibial plateaus. Inversion or eversion and rotation of the foot in the presence of a transmitted load through the heel may also result in ankle fractures. Thus for some of the occupants seated in the mid section of the aircraft some of the lower leg injuries can be explained by mechanisms other than flailing of the lower leg under the seat in front.

6.5 Summary

In this chapter a number of alternative and some novel mechanisms for the cause of injury to the pelvis and lower limbs in occupants involved in an impact aircrash have been described. It has been demonstrated that the mechanisms of these injuries are not necessarily the same as those described in the automobile industry. Indeed for those occupants seated in the regions of the aircraft where occupant protection systems were not disrupted, axial loading of the femur was shown not to be a primary mechanism of femoral fractures, knee injuries and hip injuries. Bending moments have clearly been responsible for many of these injuries. However it is probable that for this type of loading to occur the lower limbs are required to flail.

The research carried out has also demonstrated that flailing of the lower limbs following an impact air crash can be prevented by placing the feet slightly behind the vertical knee axis. This placement although preventing flailing of the lower legs results in increased axial loading of the tibia which may result in tibial plateau fractures or fractures of the os calcis.

The mechanisms identified have relied on clinical observations and measurements. These proposed mechanisms have been corroborated by findings from carefully conducted impact testing. However it is apparent that impact testing has only been able to simulate the magnitude of G in the horizontal (-Gx) plane. The effects of multidirectional vectors, and accelerations on the impact biomechanics requires the use of more sophisticated methods. In particular the effect of the vertical component of acceleration on the biomechanics requires further investigation as will be described in Chapters 7 and 8.

Chapter 7

Validation of Mathematical Computer Occupant Models

7.1 Introduction

The impact of the Boeing 737-400 aircraft resulted in a multiple vector acceleration impact. Peak accelerations of approximately 15-20G in a horizontal -Gx direction and 20-25G in the vertical +Gz vector have been identified in the mid section of the aircraft fuselage (Sadeghi et al 1989, AAIB 1990) as occurring at differing times in the impact sequence. There was no significant lateral component to the impact vector. Unfortunately impact test facilities are unable to simulate such variation in impact vectors (Chandler 1971, 1987, 1990). The experimental protocol for the impact testing (Chapter 5) thus only simulated the acceleration change in the initial -Gx vector lasting approximately 100 milliseconds.

However as a result of the development of mathematical computer models in particular in the automobile industry, computer models are available that can simulate a body's response to injury producing conditions. The theoretical advantage of these models is their ability to simulate a multi-directional acceleration crash pulse. Thus the additional effects of vertical accelerations, alterations in pitch and yaw can be simulated using such models, and will be further considered in Chapter 8.

The development of a computer simulation of passengers involved in an impact aircrash was carried out by HW Structures, Leamington Spa, Warwick, a computer analysis consultancy with an interest in the automobile industry. The occupant kinematics were investigated using a whole body model, MADYMO. This was the first time that impact simulations using computers had been employed during any aircraft accident investigation (Trimble 1991). Civil Aviation authority paper 90012 produced on behalf of HW Structures and the NLDB Study Group (1990) outlines the development of the computer simulation of occupant modelling in aircraft crash conditions. Wallace and Rowles (1990) demonstrated some of the abilities of the model by demonstrating that clinical observations could be predicted by the occupant simulation.

Criticism has however been levelled on computer models. Pitfalls include lack of validation, over sophistication, and lack of properties of biological tissues to go into the models (Panjabi 1979, Ward and Nagendra 1985). It has been stated that validation relies on correlation with experimental tests. If the models predicted response comes close to the measured results, the model is assumed to be validated (Kasarian and Von Gierke 1978, Ward and Nagendra 1985, Laananen 1985). Further it is also true to say if the models predicative response comes close to clinical observations of victims involved in an impact aircrash the model is validated.

This chapter will consider the validation of the occupant mathematical model as developed by HW Structures, using data generated from impact testing using a linear decelerator track, as described in Chapter 5. If the simulation developed is thus validated, the occupant model can be used to generate further quantitative measures, and confirm observations made as a result of the impact testing experiments and clinical review.

7.2 Aim

To demonstrate a correlation between sled tests, simulating occupants seated in forward facing seats, and the occupant model developed by HW Structures. Three different occupant seating configurations were correlated.

7.3 Method

The development of the occupant model by HW Structures is reviewed in CAA paper 90012 (HW Structures and NLDB Study Group 1990). In order to create a model of the impact environment the following data was required: seat profile measurement, seat base and stiffness measurement, seat anterior lateral spar load deflection, moment of inertia of the seat back, lap restraint data (including force link length and belt length), seat surface friction and floor foot friction.

The mathematical model was designed to be as close as possible to the impact test fixture outlined in Chapter 5. Two rows of seats were modelled with an instrumented Hybrid III dummy seated in the rear seat and a further dummy in the forward seat (H W Structures 1991).

Impact acceleration time histories for the sled tests under investigation were supplied and included the crash pulse of the impacts, pelvic accelerometer recordings and lap strap data. In addition data obtained from an accelerometer placed in the dummies head was used measuring acceleration in the -Gx plane.

The three test conditions analysed were:-

i) Brace, leg back

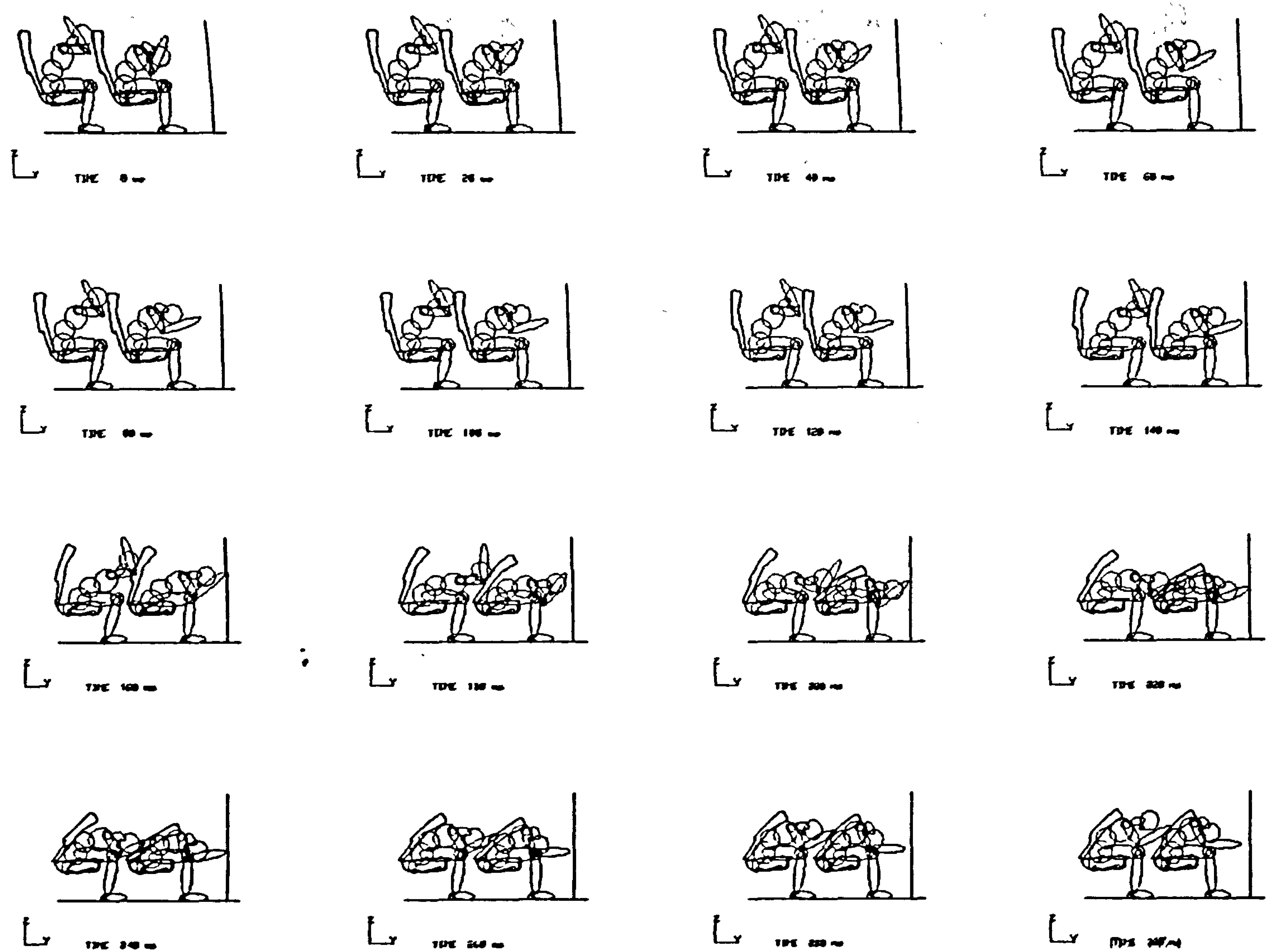
ii) Upright, leg forward

iii) Brace, leg forward

These position were analogous with those of the impact testing experiments (see figures 5.2.4, 5.2.5 and 5.2.6). The initial lap belt tightness was fixed at 20lb. The crash pulse used was the acceleration time history obtained on the 20G runs.

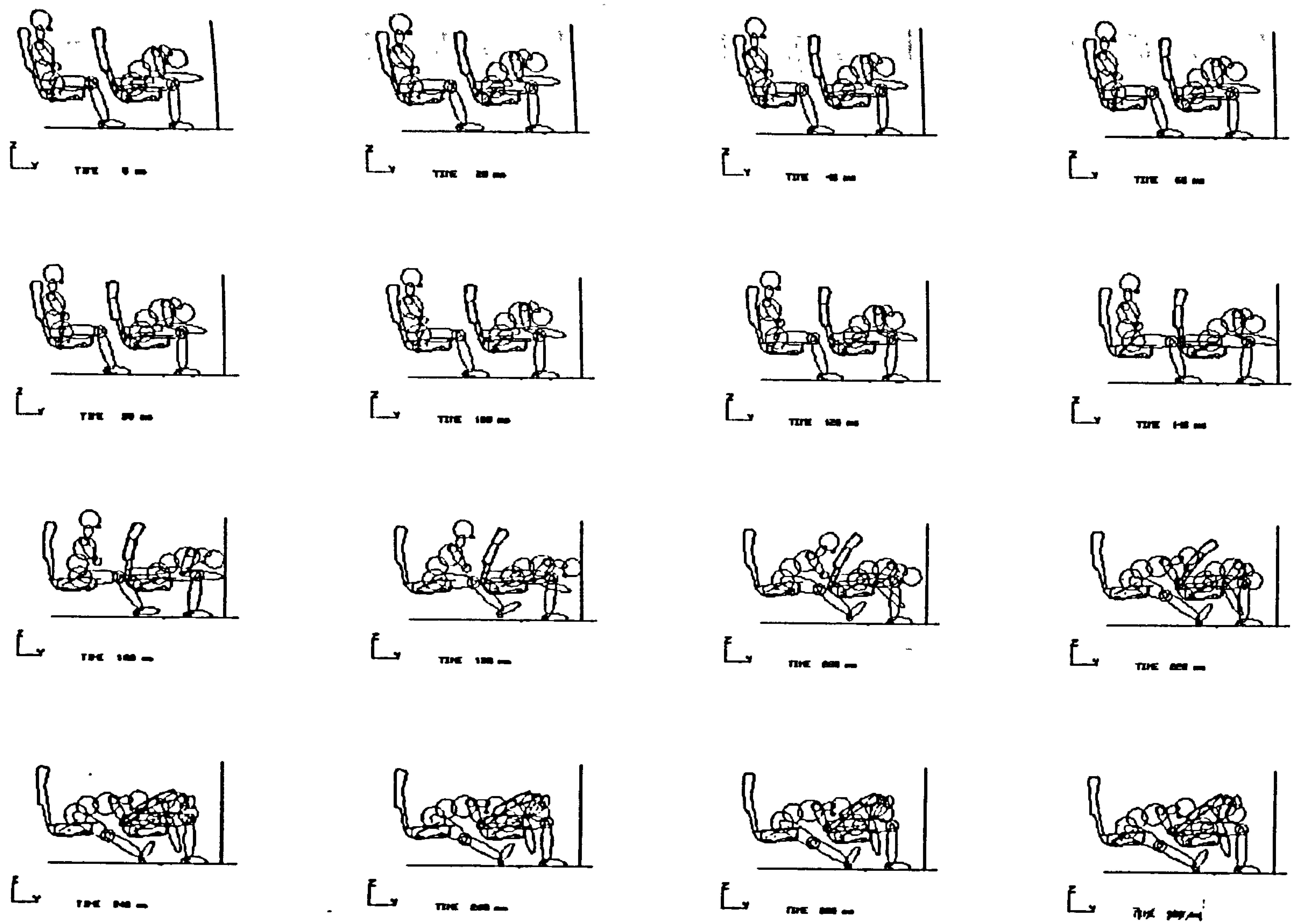
7.4 Results

The kinematic behaviour of the computer occupant models are demonstrated in figure 7.4.1, 7.4.2 and 7.4.3. as time lapse (20ms) two dimensional plots. For the braced legs



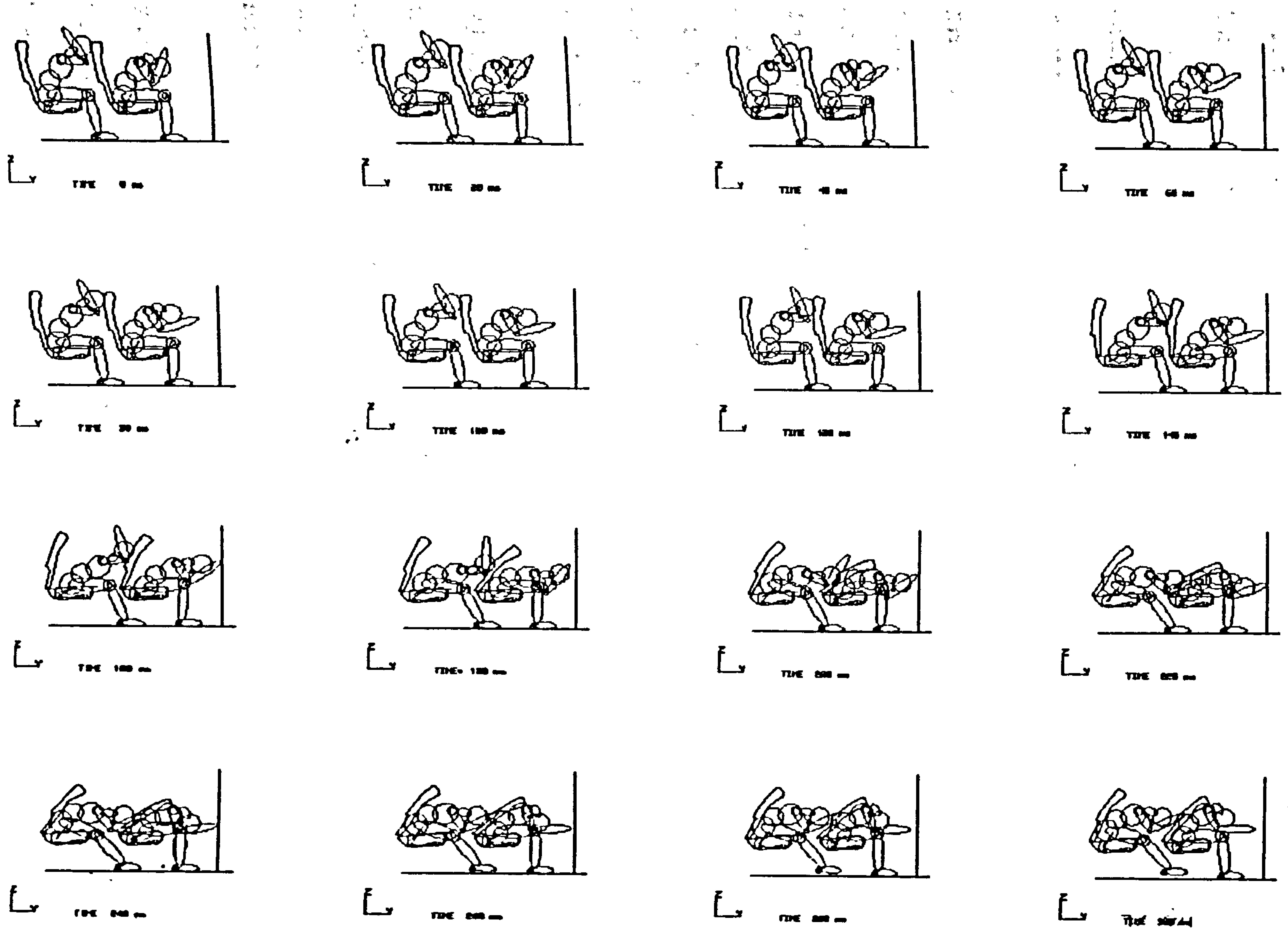
Braced Position Kinematics – leg back

Figure 7.4.1



Upright Position Kinematics – leg forward

Figure 7.4.2



Braced Position Kinematics – leg forward

Figure 7.4.3

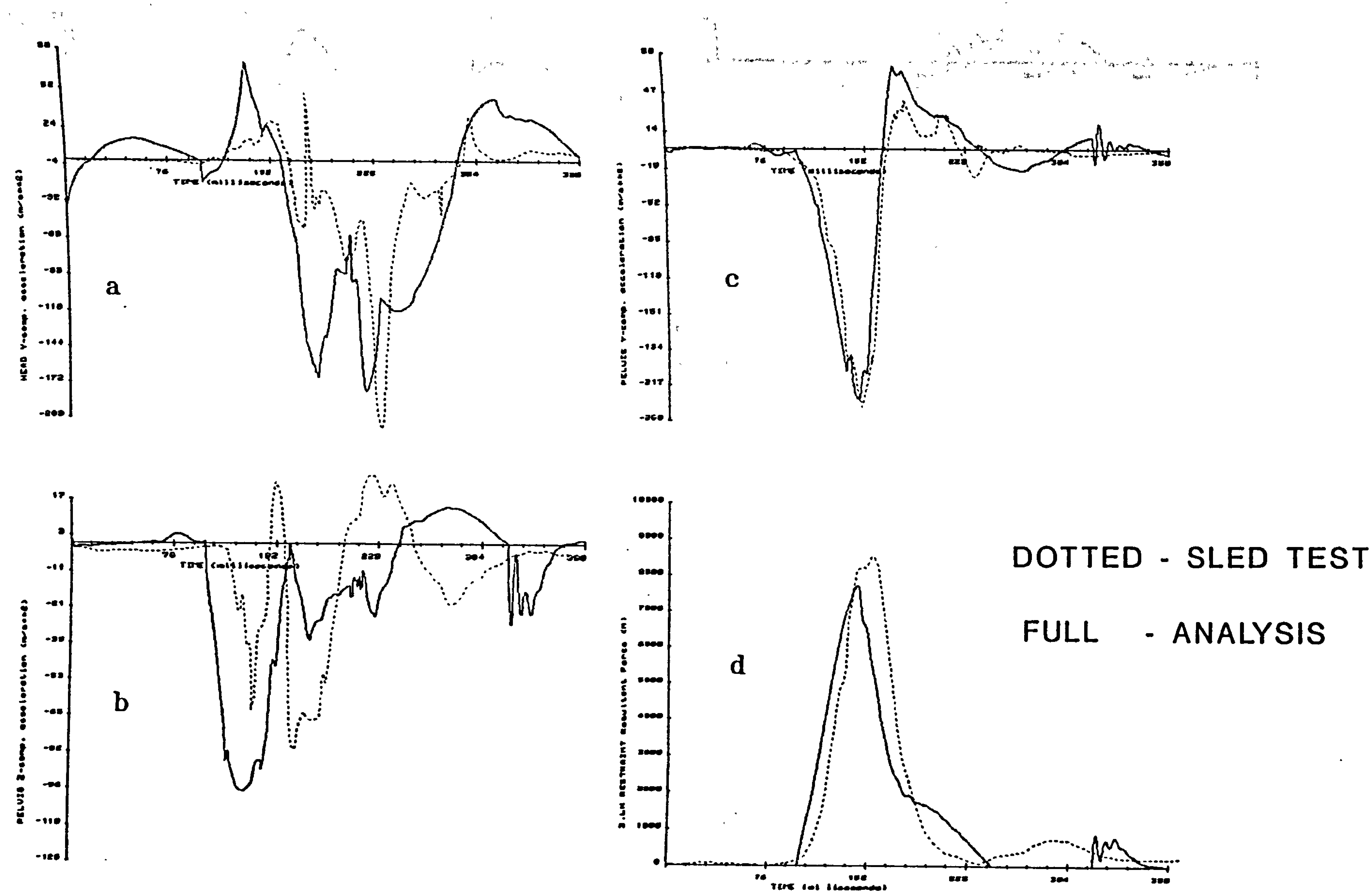
back simulation (figure 7.4.1) it can be seen that on impact the rear seated test dummy moves forward until the slack in the lap belt is taken up. The test dummy can then be seen to flex or jack knife around the belt, with breakover of the seat back of the seat infront. However it can also be seen that the lower limbs do not flail and contact of the knees with the seat ahead does not occur.

In the upright, legs forward position (figure 7.4.2) similar behaviour is seen. However flailing of the torso and head is more violent. Also and in contrast the lower limbs can be seen to flail forwards under the seat ahead, with extension of the knees. For the final test situation of braced, legs forward (figure 7.4.3) the lower limbs can be seen to move forward planted on the floor but do not flail. Knee contact is not seen.

Figures 7.4.4, 7.4.5 and 7.4.6 compare data generated from the computer model with results from impact sled tests. Graphs labelled as; a) demonstrate head horizontal accelerations, b) pelvic vertical acceleration, c) pelvic horizontal accelerations and d) lap belt forces.

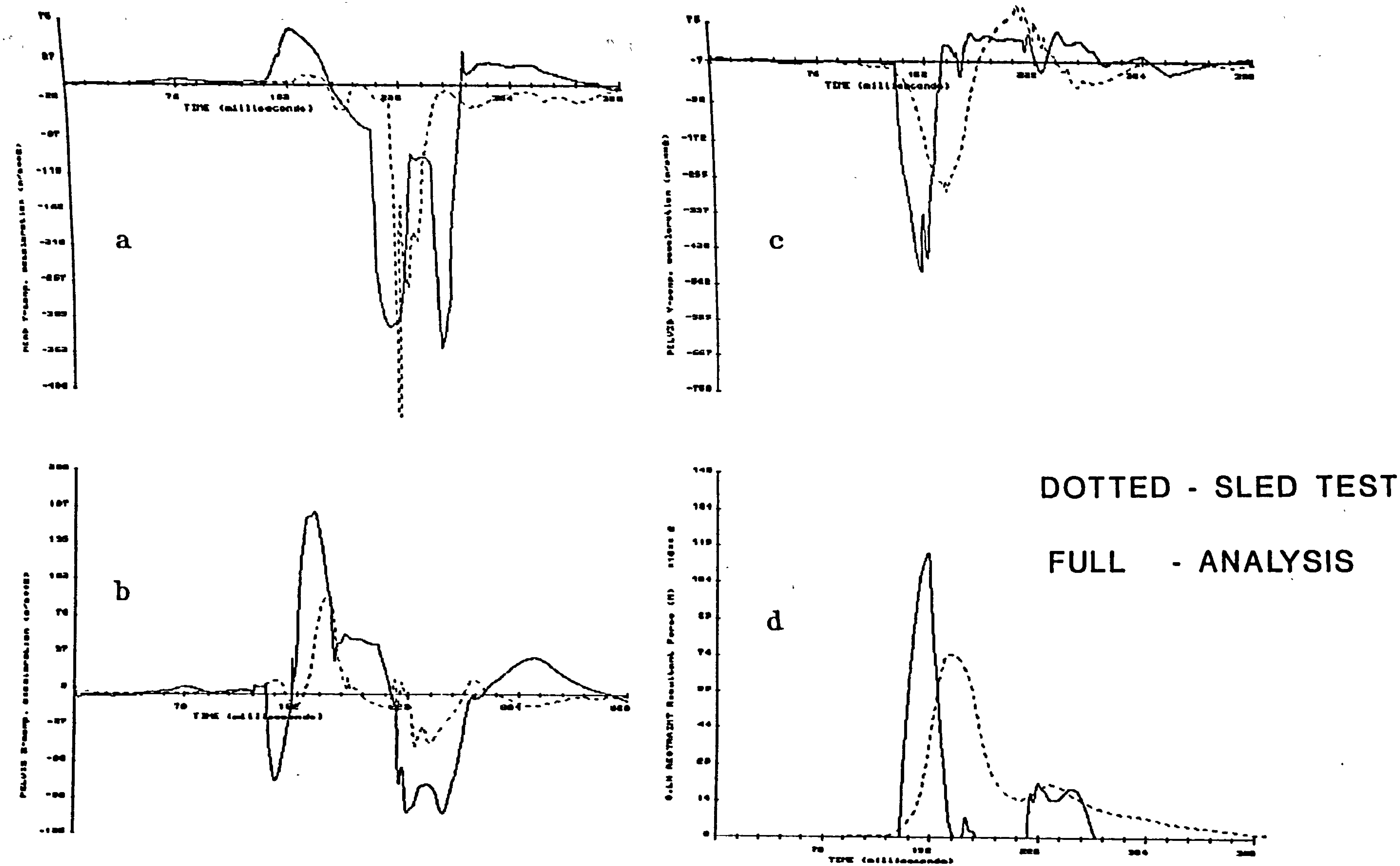
7.5 Discussion

The kinematic plots for the computer simulation demonstrate many of the features identified in impact testing. The mathematical model was able to demonstrate flail and non



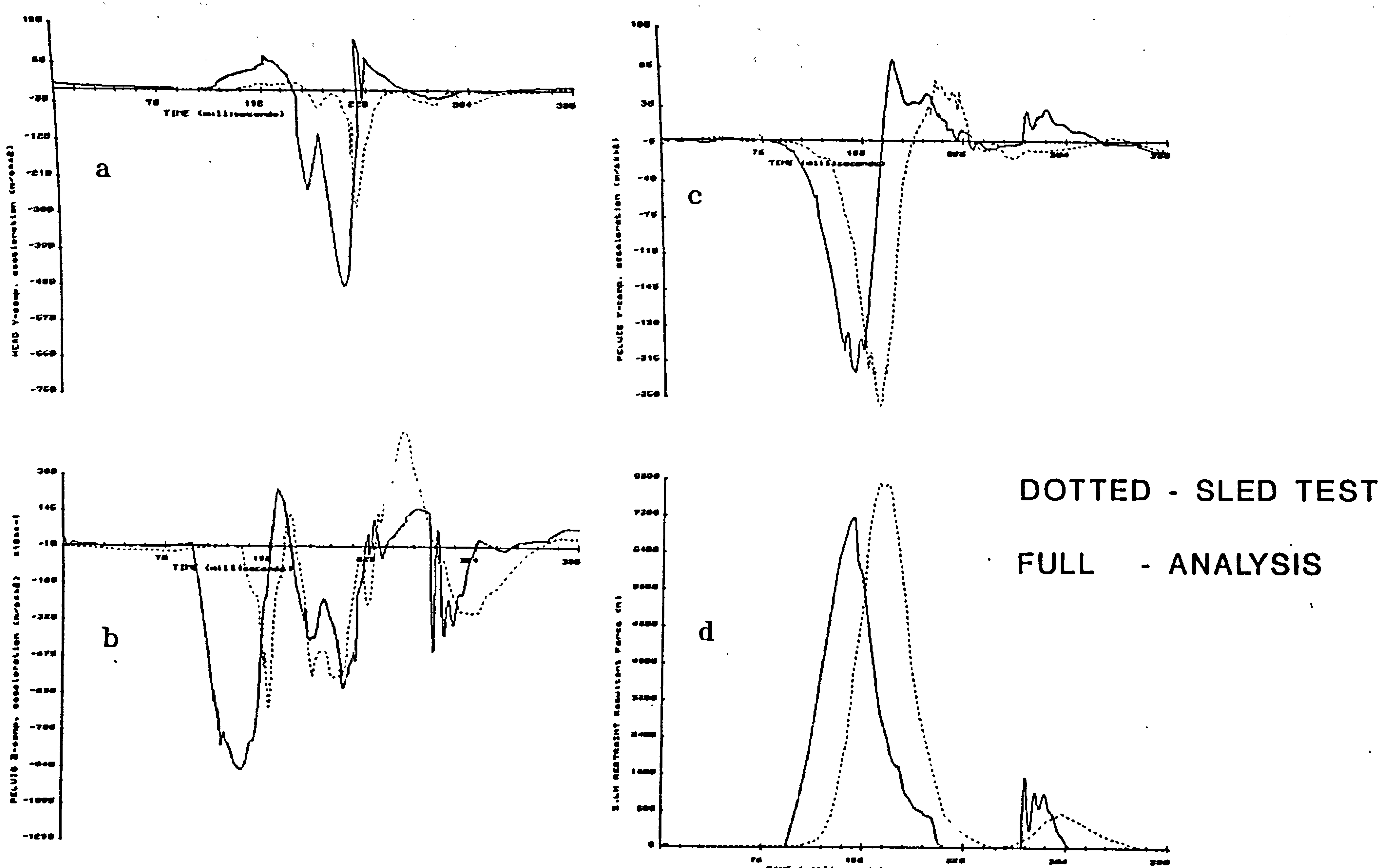
Braced Position – leg back

Figure 7.4.4



Upright Position – leg forward

Figure 7.4.5



Braced Position – leg forward

Figure 7.4.6

flail behaviour, lack of knee contact, extension of the knee with flailing and the general kinematic projection. In the situation of braced legs forward impact testing revealed a variable response, mathematical modelling demonstrated a lack of flailing behaviour. In this the computer model demonstrated the feet sliding forwards, planted on the floor.

The analysis gave excellent correlation for the pelvic horizontal accelerations and resultant belt forces and was able to predict the magnitude of the changes and the alteration with time. In the case of the head horizontal accelerations and pelvic vertical accelerations the correlation seen was less successful. However in most cases the simulation was able to predict the magnitude of the effect and the general acceleration time histories.

The reasons for possible discrepancies are numerous and are outlined in HW Structures report 5403 (1991). Possible errors include the positioning of the dummy's upper limbs, errors resulting from taping the hands together across the head in the impact tests (this would effect the kinematic behaviour of the head in particular) and failure to produce the correct contact environments in the computer model.

7.6 Conclusions

The results indicate that the computer simulation was able to demonstrate the kinematic behaviour of the dummy. In addition the model was able to predict loads and acceleration time histories for the pelvis and lap belt.

Chapter 8

Occupant Modelling of the Boeing 737–400 Aircrash

8.1 Introduction

A computer mathematical model having been validated against a known situation can be used to predict behaviour in an unknown situation (Panjabi 1979). The unknown situation in this case was the crash pulse derived by Cranfield Institute of Technology (Sadeghi et al. 1989) for the middle section of the aircraft (G-OBME) (figure 3.21.3).

From the clinical review and the impact testing experiments mechanisms of lower limb injuries have been proposed. It has also been suggested that loading of the femur over the anterior lateral spar of the occupants seat can be modified by preventing flailing and vertical knee displacement. Impact testing has suggested this may also be decreased by preventing rotation of the pelvis, possibly by means of upper torso restraint. Positioning of the lower limbs can prevent flailing and altering the characteristics of the seat squab and front of the seat may alter the loading of the femur over the anterior longitudinal spar of the seat.

However in order to confirm these effects the loads generated in the lower limbs, as a result of adopting the brace positions, needs to be known. The occupant simulation can then be used as a research tool to further investigate factors that may alter the outcome of injuries in an impact aircraft accident. A brief review of this work is outlined but a full report can be found in CAA Paper 90012 (1990).

8.2 Review of Results

Two crash brace positions were examined i) braced, feet back (figure 8.2.1) and ii) unbraced feet forward (figure 8.2.2). The kinematic plots demonstrate the behaviour of the occupant using the acceleration time history generated from the M1 Kegworth aircrash. From the kinematic plots it can be seen that significant knee contact with the seat ahead does not occur. Positioning of the lower limbs such that the foot is placed slightly behind the vertical knee axis has prevented flailing of the lower limbs in the braced position. In addition hyper-extension of the knee was identified with flailing and loading of the thigh over the anterior longitudinal spar of the seat.

Quantitative assessment of the loads experienced by the lower limb using the occupant model and crash pulse from the mid section of G-OBME is outlined in table 8.2.3.

8.3 Discussion

Using the crash pulse for the mid region of G-OBME, generated as a result of the work carried out at Cranfield Impact Centre (Sadeghi 1989, 1990) and the occupant model, developed by HW structures further quantitative information has been made available.

Brace Position Kinematics

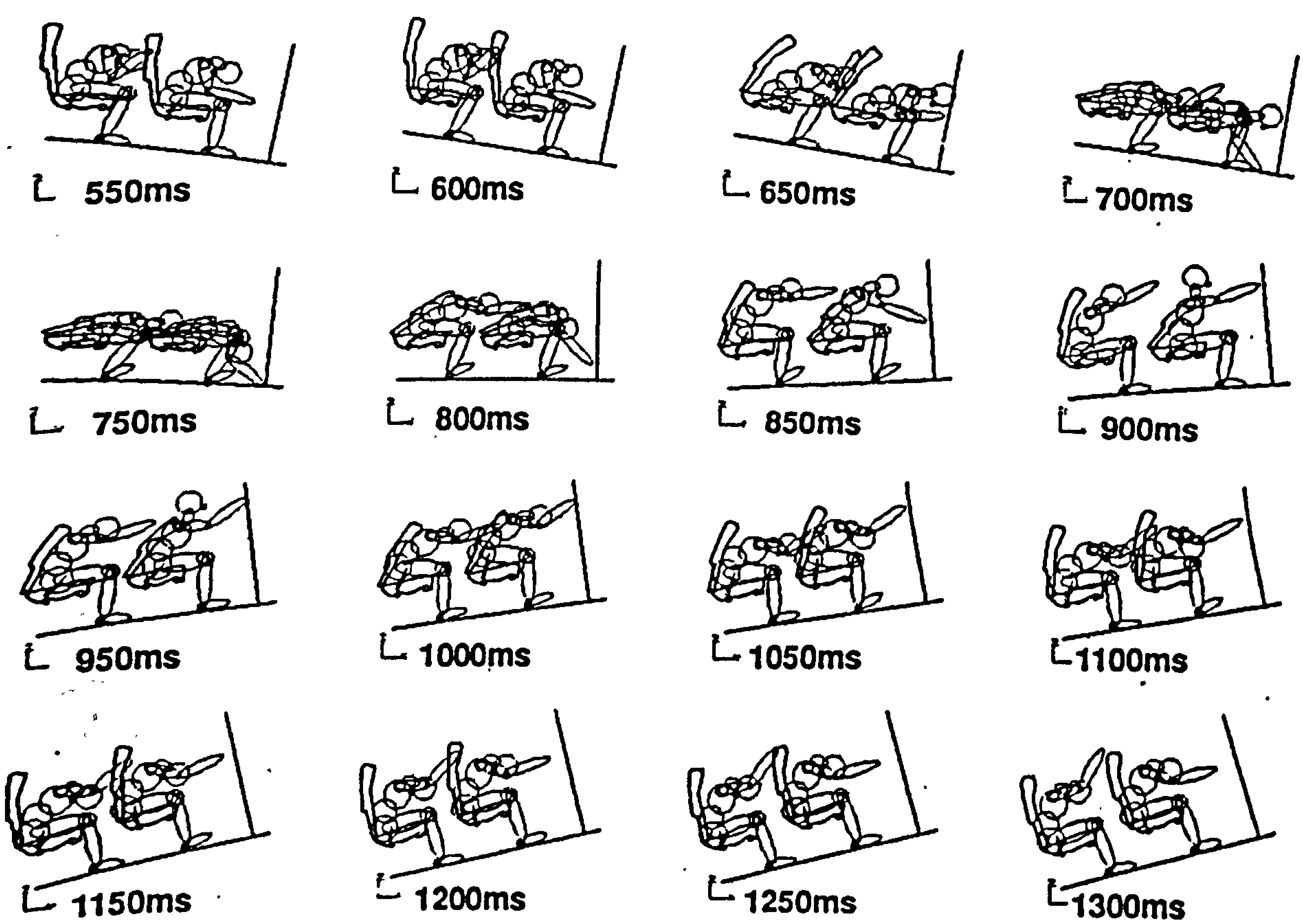
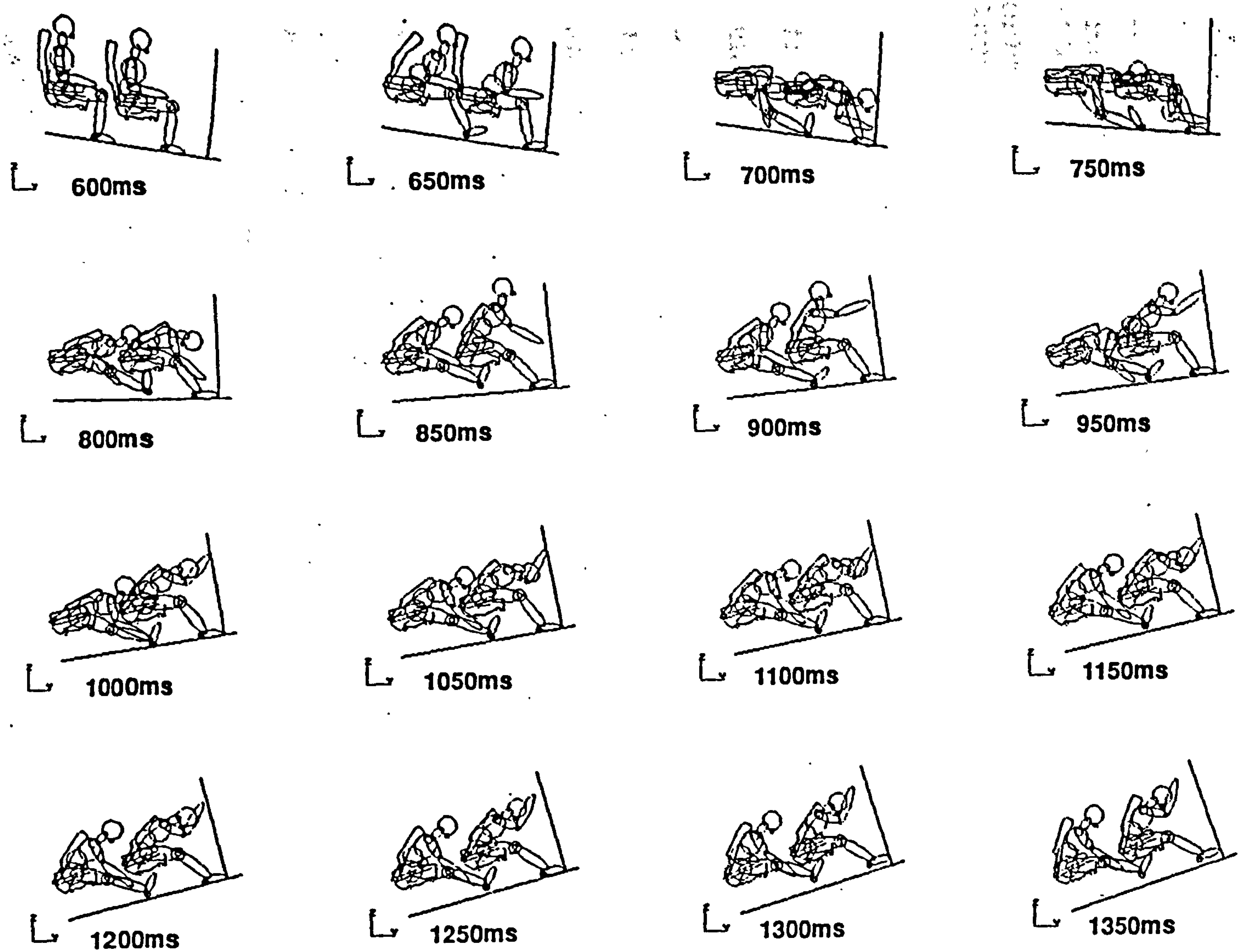


Figure 8.2.1



Unbraced Position Kinematics

Figure 8.2.2

Table 8.2.3

	Femur axial load (N)	Femur vertical load (N)	Belt load (N)	Pelvis load (N)	*Tibia load (N)	*Foot load (N)
Braced legs back	2330	2720	9441	5394	0	0
Unbraced legs forward	3367 (+45%)	1342 (-51%)	8798 (-7%)	8024 (+49%)	1152	930

*This applies only to contact loads.

In both conditions femoral axial load is low and well below the recommended injury tolerance for axial femoral loading of 10kN. The occupant modelled confirmed that significant loading of the femur as a result of contact with the seat ahead did not occur.

The femoral vertical loads generated were found to be elevated in the braced position, legs back. In this situation flailing was not seen. The femoral vertical loads are increased in non flailing situations as a result of forces transmitted through the lower leg.

The belt loads generated were found to be higher in the braced simulations with a load of 9.4kN. This was marginally greater than the unbraced position. Corresponding readings in the impact sled test using a 20G

+Gx pulse, were 8.9kN for the braced and 7.7kN for the unbraced. This suggests that the additional effect of Gz accelerations on belt loads is small.

Pelvic loads were found to be of greater magnitude in the unbraced computer simulation. Again drawing parallels with impact testing greater pelvic 'G' levels were recorded in the unbraced test situation. From clinical findings three occupants seated in the mid section, who remained seated upright on impact, all sustained pelvic injuries. This may also indicate a greater axial loading of the spine in an upright individual. The injury criteria for pelvic loading is 10kN (King 1985). Thus adopting an upright posture increases the chance of sustaining a pelvic fracture as a result of greater transmission of vertical loads.

From the tibial loads it can be seen that in the absence of flailing no contact of the lower leg occurs with the seat ahead and therefore no loads can be generated. It is known the flailing of the lower limbs into the seat ahead is a potent cause of injuries to the shins, ankles and feet. However it must be remembered that an axial load will be generated in the situations of non flail, with forces being transmitted through a planted foot. As a result of axial loading through a planted foot other injuries may be seen.

Clinical review of occupants indicate injuries previously associated with axial loading of the femur were more common in braced occupants. Impact sled testing revealed knee shear readings were found to be higher in the braced test runs and also those with the legs placed in a forward position. Unfortunately computer generated femoral horizontal and vertical loads are difficult to interpret in terms of the proposed injury mechanism of bending failure of the femur. Perhaps more relevant to the mechanisms of the injuries proposed are the bending moments created around the femur and tibia, and the axial load transmitted through the lower leg.

Unfortunately these figures are not available from the occupant simulation of the M1 Kegworth aircrash. However further developmental work using the occupant simulation model has investigated these parameters in an unbraced occupant legs forward, as part of the protocol of a wider study (Personal communication HW structures, 1991). Preliminary results for a 50% Hybrid III data set, using the crash pulse generated for the mid section of G-OBME, are indicated in table 8.3.1.

Weber (1856) demonstrated the static loads required to cause bending failure of the femur as being 233 N-m in the male and 182 N-m in the female femur at a support distance of 18.3 cm. This information indicates that the bending moment is sufficient to cause femoral fractures as a result of three point loading.

Table 8.3.1

Femur axial load	2683 N
Femur vertical load	1378 N
Femur bending moment	427 N-m
Tibial contact load	0
Heel floor load	5352 N
Belt load	9252 N

Axial loading of the calcaneum with loads of 5.5Kn have demonstrated fractures of the calcaneum (Nyquist and King 1985, Melvin and Evans 1985, Nyquist 1986). It is apparent from the above data that axial loads transmitted through the heel are therefore on the threshold of injury production.

8.4 Conclusions

Occupant modelling of the M1 Kegworth accident has been able to confirm the findings and predict mechanisms of pelvic and lower limb injuries proposed in earlier chapters. Computer generated quantitative measures indicate loads sufficient to cause injuries by the mechanisms proposed.

The model has demonstrated clinical observations of lack of knee contact with the seat ahead and also that flailing of the lower limbs can be prevented. In the absence of flailing axially transmitted loads through a planted foot are sufficient to cause injury.

The model has also indicated that femoral bending moments created are of significant magnitude and sufficient to cause femoral fracture as a result of three point loading.

The occupant model has demonstrated its ability to simulate occupant behaviour in an accident situation as well as confirm findings based on a clinical review of those occupants involved in that accident, and impact sled testing. The mechanisms of injury identified for the pelvis and lower limb may be applicable to other crash situations. The model did not assess the effects of floor distortion or the effects of fuselage fracture or loose cabin fitments in this simulation. It is hoped that as the computer model is developed and refined local structural failure might be included.

Chapter 9

The Overall Implications of the Study

The advent of crash research has seen the development of experimental equipment, advances in classification of injury and a better an understanding of the injury process. This has lead to the ability to intervene in the accident sequence to modify accident kinematics and prevent injury to occupants. As a result the effects of an accident are no longer being explained away as "Luck," "Chance," or "Fate", and attributed to "factors beyond our control".

In the aviation industry there is a genuine interest in reducing the risk of injury to aircraft occupants in the event of an accident. It is now recognised that occupants onboard an aircraft involved in an accident are likely to sustain injuries as a result of interactions with their environments: "If fire had occurred all but one survivor would have died" (NTSB 1984). "Occupants are surviving the higher crash forces, but are receiving fatal impact injuries caused by loss of restraint, or are dying needlessly in post crash fire because of injury or entrapment preclude escape" (NTSB 1984). "While leg injuries alone may not be fatal, passengers may be temporarily incapacitated to the extent rapid evacuation of the airplane is not possible" (FAA 1988). "Leg injury is of concern because of the need of rapid evacuation after impact" (FAA 1989).

Aircraft accidents are rare events and commercial flight remains one of the safest means of travel with 0.06 deaths per 100 million revenue passenger kilometres, and only 1.5 fatal accidents per million departures (Learmount 1990 (a)). A world total of 373 crashes with 10,582 deaths in commercial aircraft accidents of all kinds was reported during the ten year period up to 1990 (Learmount 1990(b)).

Despite the general trend in improved air flight safety in the 1970's and early 1980's aircraft safety seems to have deteriorated in the later part of that decade. It has been suggested that the improvement in the 1970's and 1980's was largely as a result of the introduction of safer wide bodied jets (Abelson et al. 1980). However whilst the incidence of fatal accidents has fallen, the risk of being killed in an aircrash has remained the same (Muir and Marrison 1989). The year 1989 proved to be a particularly bad year. Excluding those aircraft that were lost as a result of terrorist action 51 accidents in passenger carrying airlines resulted in 1,450 fatalities (Flight International 1990, Learmount 1990(a), 1990(b)). The major cause of aircraft crashes continues to be pilot error with some 80% of crashes analysed citing pilot error as the primary cause.

The majority of aircraft accidents usually result in only minor injuries or most commonly no injuries to the occupants (White 1966). However aircraft accidents may also result in high mortality rates (Mason 1962 1973, Lane 1975, Hill 1984, Clark 1987, Rutherford 1989, Flight International 1989, Learmount 1990). The common causes of fatalities in aircraft accidents are head and chest injuries, in that order (Mason 1962 1973, Gilles 1965, Steven 1970, Hill 1984).

In 1964 91% of the people involved in air carrier accidents received minor or no injuries, 1% received serious injuries and 8% fatal injuries (White 1966) . Lane and Brown (1975) reviewed airline statistics for the period 1960 to 1970. They found that it was unlikely that the number of seriously injured would exceed 25% of the occupant capacity of the largest aircraft. In only 1 in 20 accidents would this number be an underestimate. It was also noted that in crashes that resulted in post-crash fires the mortality rate increased from an average of 14% to an average of 34%. Rutherford (1989) reviewed civil aircrashes between 1977 and 1986 and observed that aircraft disasters resulting in more than 50 severely injured survivors occurred only three times each decade.

It is well recognised that following an aircrash a post-crash fire commonly occurs. Hill (1984) concluded that 39% of aircrashes, in the series of aircraft accidents which he reviewed and which resulted in fatal injuries to one or more occupants (Halton series), had been accompanied by a post crash fire. Clarke (1987) in his series of 537 air accidents found 128 accidents resulted in post crash fire (24%). His analysis of the fatalities suggested that 14% of the fatally injured occupants could have survived had there been no fire and that 26% of the seriously injured occupants in accidents associated with a fire, could have been injured less. Mason in 1973 discussed foot injuries in the 1967 Stockport accident in which post-mortems showed that most people died because of fire as they were unable to leave the aircraft as a consequence of their lower limb injuries. Indeed the FAA in an advisory circular (1988) indicated that leg injury is of concern because of the need for rapid evacuation after impact.

The M1 aircraft accident, on January 8 1989 afforded a unique opportunity to investigate impact biomechanics. The accident represented a major impact with no post-crash fire and substantial numbers of survivors. Modern investigative techniques have been applied to the investigation of this accident, bringing together specialists from many fields including aviators, experts in biomechanics, engineers and doctors. As a result of this co-operation different methods have been employed in the investigation of how the injuries to the pelvis and lower limbs were sustained. These methods include clinical review with classification of injuries, sled impact testing and mathematical computer simulations.

In drawing conclusions regarding the possible injury mechanisms for the pelvis and lower limbs, all reconstruction speculation was based on the mechanics known to cause the injuries sustained in aircraft accidents (Kreft 1971) and conclusions were not influenced by other impact scenarios. A body on impact will resist any sudden acceleration with a force equal to its own mass and velocity. If the force of an abrupt deceleration, during the crash sequence exceeds the physical power of resistance and the strength of the restraint system and seat, the bodies of the passengers will be hurled in the corresponding direction. By being flung against structures the occupants may sustain clearly defined or undefined injuries (Krefft 1971).

In order to determine injury mechanisms the locations and positions of the occupants involved in the aircraft accident at the instant of the crash must be known. Not only the direction in which the person was sitting but also the attitude of the head, body and extremities. Other important factors include whether the person was in an upright or crouching position at the time of the crash and whether he or she was holding on to or leaning on some support. These factors have been considered to be important in the occupants seated in the intact regions of G-OBME.

Only by identifying and defining mechanisms of injury can effective injury protection systems be designed to prevent those injuries from occurring. The design of injury protection systems should not cause an occupant to sustain injuries additional to those for which these protection systems were designed to prevent.

From the work of this thesis it is apparent that positioning of the body has implications for the type of pelvic and lower limb injuries sustained by the occupants seated in the over wing area of G-OBME. The adoption of an upright posture although protecting perhaps from femoral injuries may have serious implications in terms of pelvic and lumbar spine injury as well as head injury. Conversely the adoption of a braced position may protect against head injury but increase the chance of sustaining a femoral fracture.

Positioning of the upper trunk and head is further complicated by the positioning of the lower limbs on impact. Flailing of the lower limbs is not an inevitable response in an impact aircrash, and appears to be prevented if the feet are placed slightly behind the knees. If this is the case for other impact scenarios then this may have implications in terms of preventing those injuries to the lower limbs seen as a result of impact with the seat in front. However if flailing is prevented the lower limb may act as a strut and prevent loading of the femur over the anterior lateral spar of the seat. But at what cost?. It may then transpire that axial loading of the planted foot may be sufficient to cause injury.

FAA 14 CFR part 25 (1988) states "femur loads should therefore be measured during the dynamic tests where leg injuries may result from contact with seats or other structure. A measured axial load of 2250 pounds (10kN) along each femur should not be exceeded during these tests. This is the same as the maximum allowed by the Federal Motor Vehicle Safety Standard No. 208." As has been demonstrated throughout this thesis, clinically, with impact sled testing and with mathematical modelling, impact of the knee with the seat in front is of little significance.

Many of the injury protection systems in the automobile industry are aimed at preventing or diminishing the effects of the instrument panel syndrome. It is apparent that in the majority of automobile accidents no vertical acceleration pulse is experienced, yet in the majority of impact aircraft accidents their is a significant vertical component. Likewise in the automobile industry the knee femur pelvis mechanism as well as lateral impacts are a potent factor in injury causation to the pelvis and lower limbs. Conversely it appears that in the M1 aircrash axial loading of the femur was not a significant factor in the causation of the pelvic and femoral injuries in occupants seated in the intact regions of G-OBME.

It has been suggested that three point loading of the femur caused by the lap belt, the anterior spar of the occupied seat and flailing of the lower limbs with impact against the posterior spar and sub seat assemblies of the seat ahead, is an important injury mechanism in impact aircraft accidents. In addition the forceful rotation or jack knifing of the torso around the lap belt has implication for injuries around the hip. It would therefore seem more appropriate to regulate for the bending moment created around the femur rather than the axial load transmitted up the femur; the prevention of jack knifing of the torso; the flailing of the lower limbs, and to improve the impact friendliness of the seat ahead.

From the previous discussion it is apparent that the anterior lateral spar of the seat is an important loading structure for the bending moment created around the femur. It may be possible to modify the design of seats to reduce the loads. Alterations in the seat squab stiffness or the position of the anterior spar may effect the biomechanics. The findings of this thesis should be confirmed for seats of other design. Modification of seating to reduce bending moments produced in the femur should also be investigated further.

Pelvic fractures have been identified in passengers onboard G-OBME analogous to those seen in occupants of automobiles involved in lateral car impacts. No significant lateral acceleration existed in this accident. A forward facing occupant restrained only by a lap belt will have considerable forces transmitted through the restraint device. It has been demonstrated that transmission of these loads are sufficient to cause pelvic injuries. Legislation should indicate an injury tolerance limit for loads transmitted through a lap belt.

As a result of the accident of Boeing 737-400 G-OBME on the 8 January, it is apparent that the mechanisms of pelvic and lower limb injury differed from the accepted philosophy of injury causation. It has been shown that the flailing behaviour of the lower limbs can be modified and that axial loading of the femur is not necessary for injuries to the femur and pelvis to occur. When the feet are placed on the floor slightly behind the vertical axis of the knee lower limb flailing is prevented and this position has been incorporated into a new crash brace position (NLDB Study Group 1990). Adopting such a position may also reduce the incidence of femoral fractures, as it prevents loading of the femur over the anterior spar of the seat. This effect requires further investigation. If flailing can be prevented an injury tolerance level should be defined for axial loading of the lower leg (tibia).

Individuals can withstand large forces but there is a limit to how the impact loads may be distributed to an occupant in order to prevent injury. Within the constraints of forward facing seats it is apparent that severe injury is possible without secondary impacts. It is also apparent that considerable loads will be transmitted through a lap belt. This being the method used to prevent an occupant from becoming a free projectile. Minor alterations in the brace position that may prevent injury at low impact loads or in different crash scenarios may result in additional injuries in other crash scenarios. It may be that with present seating arrangements injuries to the pelvis and lower limbs may be difficult to prevent. The problems of developing an effective crash brace position and occupant safety systems is thus not easy.

Previous aircraft accident investigations have not undertaken such thorough documentation and classification of the injuries to both the survivors and the non-survivors, and have not looked at a detailed analysis of the position adopted by each occupant at the time of impact. Hill (1984) has stated that of the "vast number of accidents which occur annually only a few are properly analysed. Volumes of useful material which might have saved lives and would certainly have provided evidence which would have reduced the level of injuries has been ignored." As a consequence there is no previous work of sufficient detail to allow comparison with other air crashes and thus to confirm the findings of this thesis.

In addition no two aircraft accidents can be considered exactly the same and therefore the findings extrapolated from one incident will not necessarily be true of other events. The National Transportation Safety Board (1981) and FAA (1988) now recognises two impact scenarios for the dynamic testing of aircraft seats. These have been derived from automobile regulatory crash testing. One can question whether these scenarios have fidelity for all crash situations in the same way as the findings of this research may not be totally applicable to other situations.

What methods exist to allow exploration of the problem of bracing for impact and occupant safety system design? Mathematical computer models of occupant kinematics have for a number of years been used to investigate injury biomechanics in the automobile industry. Clearly this technique is attractive in that it can simulate differing crash scenarios, model for individual occupants in terms of both individual variation and position adopted at impact, as well as investigate alterations in safety systems design. However in order to develop an effective model you require good information.

It is hoped that in the future aircraft accidents will be investigated in a similar manner to that outlined in this thesis. Although it may prove impossible to prevent accidents from happening it is possible by carrying out meticulous studies on those aircraft which crash, to influence the accident sequence in such away as to reduce the future risk of injury to people involved in an impact aircraft accident.

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Appendix 1

The Nottingham, Leicester, Derby, Belfast Study Group

As a consequence of the aircraft accident of Boeing 737-400, G-OBME, on the 8 January 1989, a unique opportunity was presented to analyse the injuries of passengers both living and dead. A study group was initiated by Professor Angus Wallace and Mr Christopher Colton of the Department of Orthopaedic and Accident Surgery in the University of Nottingham Medical School. The project had the following aims:

1. To identify and document all soft tissue and bony injuries to crash victims.
2. To analyse the mechanical forces generated by the crash at each seat position.
3. To investigate the likely cause of the injuries sustained.
4. To document the immediate and definitive management of all injuries sustained.
5. To follow-up all injured patients for six months and establish the outcome.
6. To identify the nature and number of missed injuries.
7. To carry out an internal audit of our own performance in looking after crash survivors.

The study group collaborated with the local coroner and with the Air Accident Investigation Branch, Farnborough. In conjunction with an Engineering Analysis Consultancy, H.W.Structures Ltd, Leamington Spa, a computer simulation of occupant kinematics of passengers involved in the Kegworth aircrash was developed.

The study group has the following members:

Chairman Professor W A Wallace MB, ChB, FRCS(Ed), FRCSEd(Orth). Professor of Orthopaedic and Accident Surgery, University of Nottingham.

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Appendix 2

Breakdown of Pelvic and Lower Limb Injuries for all Occupants of G-OBME

The following tables list all pelvic and lower limb injuries identified in the occupants of G-OBME. Long bone fractures have been classified according to the 'A.O. fracture classification' (Muller, Nazarian & Kock 1987). This classification sorts fractures into types which are easily identifiable. The fracture types can also be related to the mechanism that produced the fracture. Open fractures have been graded according to Gustilo and Anderson, 1976 classification. Fractures of the talus have been recorded according to Hawkins classification (1970).

For the purpose of the analysis survivors were those occupants who survived the accident to be admitted to a hospital ward (83 patients). These individuals had their injuries well documented with X-rays, clinical notes and other investigations. The remaining occupants were categorised within the non-surviving group (n=43) and include four occupants that survived the impact but died soon after removal from the wreckage.

Classification of fractures in the non-survivors could not be accurately undertaken as X-rays were not available. It was also likely that some fractures were not identified at necroscopy. Further post-mortem descriptions of compound injuries were insufficient for accurate classification although it was apparent that many were of a severe degree.

Injuries to the pelvis and lower limbs have been categorised according to anatomical location into the following groups:- Pelvis, Femoral, Knee, Tibia, Ankle and Foot.

Key to tables

Patient number:- This refers to the number allocated to occupants on board the aircraft in order to prevent identification of individuals. The number of patients sustaining injuries to each region is recorded at the end of each table.

Fracture complex:- Injuries to a region may consist of more than one fracture type. This column records the number of fractures present within the bone. The fracture types are recorded at the end of each column.

Compound fracture and grade:- This records the side (if not apparent) and grade of injury. Compound injuries that are not classified are recorded as 'C'. The number of compound injuries for each region is recorded at the end of each table.

ISS:- The Injury Severity Score for all patients is recorded and the average ISS for each region demonstrated.

Abbreviations used

DISLOC	- dislocation
#	- fracture
H	- Hawkins classification
INF	- inferior
LAT	- lateral
L	- left
MT	- Metatarsal
PCL	- posterior cruciate ligament
PP	- proximal phalanx
R	- right
SI JT	- Sacroiliac joint
SUP	- superior
T/F	- tibia/fibula

PELVIC INJURIES - SURVIVORS

PATIENT NUMBER	ISS	TYPE OF INJURY
7	43	#R INFERIOR PUBIC RAMUS
38	34	POST. DISLOC R HIP
28	22	POST. #/DISLOC R HIP
61	17	#R ILIUM+R SUP & INF PUBIC RAMI
24	45	#R ILIUM+ BILATERAL SUP PUBIC RAMI
73	14	#L ILIUM
46	27	#R & L ACETABULUM
79	14	#R SUP & INF PUBIC RAMI
104	12	#L ACETABULUM
2	6	#R ILIUM
29	19	#/DISLOC L HIP
60	12	BILATERAL # ILIAC CREST
42	9	SUP & INF PUBIC RAMI
52	17	#/ DISLOC ACETABULUM
111	10	#R ACETABULUM
19	14	#R ACETABULUM
62	4	#BILAT SUP & INF PUBIC RAMI
90	27	DIASTASIS SI JT & SUP PUBIC RAMI
84	10	#BILATERAL SUP PUBIC RAMI
95	5	DIASTASIS SYMPHIS PUBIS
77	5	DISLOCATION R HIP
83	19	DIASTASIS R SI JT & RUPTURED BLADDER
54	5	DISLOCATION L HIP
--	--	
23	Average = 16	

PELVIC INJURIES - NON-SURVIVORS

PATIENT NUMBER	ISS	TYPE OF INJURY
68	75	#L PUBIC RAMI
96	75	BILATERAL SI JT DISLO & DIASTASIS SYMPHIS PUBIS
1	45	DIASTASIS SYMPHIS PUBIS
76	33	#R ILIAC BLADE
67	57	#BILAT PUBIC RAMI & DIASTASIS SYMPHIS
124	75	DIASTASIS SYMPHIS PUBIS
16	75	DIASTASIS SYMPHIS PUBIS
9	75	BILATERAL SI JT DISLOC & SYMPHIS DIASTASIS OF SYMPHSIS
8	75	BILATERAL SI JT DISLOC & DIASTASIS SYMPHSIS PUBIS
--	--	
9 Average =	65	

FEMORAL FRACTURES - SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	COMPOUND # and GRADE	ISS	TYPE(AO)
43	1		10	L31A2
100	1		22	L31A3
114	1		14	L31A2
74	2		14	L31A1/L32A3/R32A3
90	2		27	L32B2/R32A2
84	2		10	L32A2/R32B1
85	1		27	L31A2
46	1		27	R32B2
79	1		14	L31A2
29	1		19	L31A2
117	1	G2	13	L32B1
20	1		11	L32C2
51	1		26	R32A3
65	1		19	L32B2
111	1		10	R32C3
41	1		27	L32B2
83	1		19	L TROCHANTER
7	1		43	R32C3
41	1		27	L32B2
--	--	--	--	
19	22	1	Ave = 20	

FEMORAL FRACTURES - NON-SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	COMPOUND # and GRADE	ISS	TYPE(AO)
70	1		26	R32
71	1		50	R32
45	1		43	R32
102	1		34	R32
118	1		34	R33
56	1		75	R32
88	2		75	R32/L33
57	1		75	L33
59	1		75	L32
13	1		75	L32
96	1		75	L32
16	1		75	L32
--	--		--	
12	13	Average =	59	

KNEE INJURIES - SURVIVORS

PATIENT NUMBER	INJURY COMPLEX	COMP.& GRADE	ISS	TYPE(AO)
99	1		14	L SUP T/F SUBLUXATION
28	1	C	22	R COMP LAC
112	1		22	R41B1
48	1		9	L EFFUSION
87	1		17	R PCL RUPTURE
46	2	C	27	R&L COMP LAC
69	1		5	L EFFUSION
10	1		9	L EFFUSION
77	1	C	5	R COMP LAC
107	1		9	L LAT LIG RUPTURE
4	1		6	R PCL RUPTURE
37	1	G3	11	R41B1
52	1	C	17	L41A1
101	1		24	L LAT LIG RUPTURE
54	1	C	5	L COMP LAC
24	1	C	45	R COMP LAC
86	1		5	R41B1
--	--	--	--	
17	18	7	Ave = 15	

KNEE INJURIES - NON-SURVIVORS

PATIENT NUMBER	INJURY COMPLEX	COMP.& GRADE	ISS	TYPE(AO)
71	1		50	L #/DISLOCATION
45	1		43	L DISLOCATION
102	1		34	R DISLOCATION
13	1	G3	75	R #/DISLOCATION
110	1		75	L DISLOCATION
---	--	--	--	
5	5	1	Ave = 55	

TIBIAL FRACTURES - SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	COMPOUND # and GRADE		ISS	TYPE(AO)
53	1	G2		38	L42B3
89	1	G2		27	R42B3
119	2	RG2 & LG2		19	R42C3/L42C3
109	1	G1		10	L42B2
99	1			14	L41A3
98	1			11	L41B2
44	1			14	R42B3
7	1	G2		43	L43C3
75	1	G2		41	R42A3
123	1	G2		50	L42B2
41	2	G3		27	L42A3/L41C2/R42B2
29	2	GL2		19	R42C3/L41C2
10	1			9	L42B2
5	1	G2		10	R42A3
122	1	G2		22	R42B2
17	1	G1		10	R42A2
37	1			11	R43C2
65	1			19	L42A2
52	1	G3		17	R43A3
83	1	G2		19	R42B2
74	2			14	R42A3/L42A3
26	1			14	R41B2
38	1	G3		34	L43B3
28	1			22	R43C3
112	1	G3		22	L42B2
46	1	G3		27	L42A2
32	1			29	R43B2
--	--	--		--	
27	31	19	Ave =	22	

TIBIAL FRACTURES - NON-SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	COMPOUND # and GRADE	ISS	TYPE(AO)
70	1	C	26	L42
71	2	CR	50	L42/R43
106	2		29	R42/L42C
118	1	C	34	L42C
8	2	C	75	R41/L42
9	2	C	75	R42/L42
55	1		75	L42
56	1	C	75	R42
15	2	CR	75	R41/L42C
16	2	CR	75	R41/L41
88	1		75	R42
22	1		66	L42
33	2	CR	75	R42C/L42
125	1	C	75	R42C
57	1	C	75	L42
58	2	CR	75	R42/L42
12	1		33	R42
124	1	C	75	L42
13	2	C	75	R42/L42
110	1	C	75	R43
116	1	C	45	R42
31	2	C	66	R41/L42
105	2	CL	75	R42/L42
68	1	C	75	R42
96	1	C	75	42 (TRAUM AMPUT'N)
39	1	C	43	R42
47	1		41	L42
--	--	--	--	
27	38	25	Ave = 63	

ANKLE INJURIES - SURVIVORS

PATIENT NUMBER	INJURY COMPLEX	COMPOUND # and GRADE	ISS	TYPE(AO)
53	2		38	R43C3/L43B2
21	2		5	R44B1/L44B1
109	1		10	L44C2
98	1	G3	11	R44A3
7	1	G3	43	R44B3
75	1		41	L44A2
26	2	GR2	14	R44C3(+TALUS)/L44A2
24	1	G2	45	R44C2
63	1		3	R SPRAIN
69	2		5	L43B2/R44C1
10	1		9	R SPRAIN
3	1		6	R SPRAIN
17	1C3		10	L44C1
126	1C3		11	R44C1
65	1	G3	19	R44C2
52	1		17	L44A1
83	1		19	R SPRAIN
101	1		24	DISLOCATION - NO #
41	1	G3	27	R44C2
87	1		17	R TALAR CHIP
84	1		10	L44B3
35	1		41	L44A1
90	1		27	L43B2
--	--	--	--	
23	26	8	Ave = 20	

ANKLE FRACTURES - NON-SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	# COMPOUND and GRADE	ISS	TYPE(AO)
91	2	C	27	R44/L44
103	2	C	21	R44/L44
45	1	C	43	L44
102	1		34	R44
40	1	C	34	L44
118	1	C	34	R44
27	2	C	29	R44/L44
8	1		75	R44
9	1	C	75	R44
55	1	C	75	R44
16	2	CL	75	R44/L44
88	1	C	75	R44
34	1		75	R44
125	1	C	75	L44
11	1	C	34	R44
105	1	C	75	R44
68	1		75	L44
96	1	C	75	L44
39	2	C	43	R44/L44
--	--	--	--	
19	24	19 Ave =	55	

FOOT INJURIES - SURVIVORS

PATIENT NUMBER	INJURY COMPLEX	COMP.& GRADE	ISS	TYPE(AO)
53	1		38	L TALAR #/DIS H3
43	1		10	R LISFRANC+ 4&5 MT
119	1		19	L LISFRANC
98	2	RG3 & LG3	11	R&L TALAR #/DIS H3
44	1	G3	14	L SUBTALAR #/DIS
32	2	GR2	29	#R 2-5MT + #R 1stMT HEAD + #L GT TOE.
95	1		5	R 2-4MT
66	2		9	R & L LISFRANC
52	1	G3	17	L LISFRANC + L 3MT
115	1	G2	5	L 3PP
75	1		41	R LISFRANC
38	1		34	L CALCANEUS
42	1		9	R LISFRANC/R TALUS# H1
65	1		19	L 2&3 MT
123	1	C3	50	R TALAR#/DIS H3
41	1		27	R LISFRANC
126	2	RG3 & LG3	11	R TALAR #/DIS H3 + L TALAR #/DIS H2
--	--	--	--	
17	22	9	Ave = 20	

FOOT FRACTURES - NON-SURVIVORS

PATIENT NUMBER	FRACTURE COMPLEX	COMP.& GRADE	ISS	TYPE(AO)
15	1	C	75	#R 1&2 TOE
88	1	C	75	R TALAR#/DIS H3
57	1	C	75	L LISFRANC
58	1		75	R LISFRANC
68	1	C	75	#R 2MT
16	1		75	L LISFRANC
--	--	--	--	
6	6	4	Ave = 75	

Appendix 3

Anthropometric Measurements made on Occupants Seated in the Mid Section of G-OBME

Anthropometric measurements were made on the survivors seated in rows 10 - 20 on board Boeing 737-400, G-OBME. Measurements were made in accordance to guide lines for measurements as stated by Bolton et al. (1974). Measurements were made using standard measuring tapes and a standard anthropometer. Thirty one of the thirty eight occupants were measured.

The following measurements (as indicated below) were made and are recorded in table Appen 3.1. The means and standard deviations have been calculated for all measurements and are displayed in table Appen 3.2.

Weight:-

Standing on spring scales.

Height:-

Standing erect head facing forward. Measurement from floor to vertex.

Head diagonal:-

Measurement made with standard anthropometer, being the distance from the chin (with the jaw closed) to the vertex so that the maximum diagonal was measured.

Head breadth:-

Measurement with standard anthropometer recording maximum breadth of head.

Cervical spine:-

Length measurement with standard tape from C7 spinous process to the base of the skull.

Sitting height:-

Sitting erect with head forward and back clear of rear wall. Measurement made from floor to highest point of vertex and then subtracting the height of the chair.

Sitting height to C7:-

Sitting erect with head facing forwards and back clear of wall. Measurement from floor to C7 spinous process with the subtraction of the chair height.

Shoulder width:-

Sitting erect with shoulders relaxed. Measurement using standard anthropometer as greatest distance between maximum prominence of deltoid muscle.

Shoulder olecranon length:-

Sitting erect with shoulders relaxed, elbows held lightly against sides. Measurement made with standard anthropometer from highest point of acromium to lower edge of left olecranon process.

Olecranon to third finger (elbow/fingertip length):-

Arm held horizontal with elbow touching wall, fingers outstretched in line with the forearm. Measurement from end of wall to tip of the third finger.

Chest circumference:-

Circumference standing erect with measurement made at level of nipples. Measurement made during quiet breathing.

Buttock/knee length:-

Sitting with buttocks firmly against chair back, thighs parallel to the ground, feet flat on floor. Measurement with standard anthropometer from anterior patella to seat back.

Buttock/sole length:-

Sitting on floor with back to the wall, legs straight in front. Measurement from wall to heel.

Floor/knee (sitting knee height):-

Sitting erect with shins vertical, feet flat on floor. Measurement from floor to upper surface of thigh above femoral condyles.

Foot length:-

Maximum distance from heel to most prominent toe using standard anthropometer.

Abdominal circumference:-

Standing erect. Measurement made at the natural waist indent with measuring tape.

Pelvic width:-

Standing erect. Measurement made with standard anthropometer as a greatest distance between greater trochanters of the femur.

Pelvic circumference:-

Standing erect. Measurement made with standard tape at the level of the greater trochanters of the femur.

Anthropometric Measurements on Occupants

Append 3.1

SEAT	SEX	AGE yrs	HEIGHT cm	WEIGHT kg	Head diagonal cm	Head breadth cm
10C	M	20	168	67	26	15
10D	M	21	185	66.7	26	15.5
10F	F	20	159	49.2	24.5	14
11A	F	56	165	95.5		
11B	F	59	148	58	23.5	15.5
11C	F	25	166	54	24	15.5
11D	M	25	169	73	25	15.5
11F	M	24	188	101		
12A	F	52	157	55.5	24	15
12B	F	32	164	63	23.5	14.5
12C	F	31	159	55	24	14
12F	M	40	182	95	23.5	16
14A	M	38	170	76.5	26	15
14D	M	18	178	55		
14C	F	26	168	59.5	23	14
14F	M	24	184	84	26	16
15A	M	38	170	68		
15B	F	38	165	52		
15C	M	29	167	69	24	15.2
15F	M	50	169	83.5	25	15.5
16A	M	65	162.5	76.5	25	15
16C	F	24	162.5	57	23.5	14
16D	F	32	161	61.4	24.5	14.5
16F	F	39	166	60.5	23	14.5
17A	M	19	183	78	26.5	15.5
17B	M	69	173	63.5		
17D	M	62	182	76	25.5	15
17F	F	36	160	54		
18A	F	26	148.5	65.5	24.3	15
18B	M	21	169.5	64.6	24.5	15
18C	M	41	170	82.5	26.5	15.5
18D	M	24	178	63.5	26	15.2
18F	M	58		DECEASED		
19A	M	23	180	67	25	15.2
19E	M	24	178	63.6	24.5	15
19F	M	21	181	75	25	15
20A	M	44	176	89		
20C	M	28	180	82.7	25	14.5

Average Anthropometric Measurements

	AGE yrs	HEIGHT cm	WEIGHT kg	Head diagonal cm	Head breadth cm
Average	34.79	170.05	69.21	24.72	15.00
Std. Deviation	14.37	9.79	13.01	1.03	0.56
Ave. (female)	32.93	175.44	74.67	25.28	15.26
Std. (female)	14.04	7.13	8.71	0.84	0.38
Ave. (male)	35.88	159.91	58.05	23.80	14.59
Std. (male)	15.04	6.68	4.59	0.54	0.58

Append 3.2

Anthropometric Measurements on Occupants

Append 3.1

C spine cm	Sit height cm	Sit - C7 cm	Shou. width cm	Shou - Olec cm	Olec - 3fin cm	Chest cm
12	92	79	44	36	47	92
15	92	80	43	37	48	88
13	82	62	41	35	44	80
8	74	56	44	35	43	105
13.5	86	63	39	35	43.5	79
16.5	92	69	45	37.5	43	99
10	81	65	40	32	45	80
12	87	64	41	37	43	90
12.5	87	76	39	32	42	88
11	93	64.6	46	37.5	50.5	109
13	64	77	42	37	48	94
12	82	65	41	36	44	92
14	96	70	48	39	51.5	95
9	86.5	77	44	36	44	92
14	90	69	47	40	48	106
11	93	82	41	37	45	103
13	87	65	41	32	42	90
12	95	82	36	34	43	89
14	83	63	41	34	43	99
9	91.5	69.5	56	40	49	103
14	96.5	70	41	36	47	100
10	83	62	34	30	40	95
14	83	66	42	37	47	93
10	88	64	44.6	34.4	45.3	104
12	89.5	62.2	42.6	35.2	47	89.5
11.5	91.8	65.5	44	37.4	47.5	64
12.5	89	65	43	39.5	47	90
13	94	84	43	35	48	86
13	91	80	43	37	45	88

Average Anthropometric Measurements

C spine cm	Sit height cm	Sit - C7 cm	Shou. width cm	Shou - Olec cm	Olec - 3fin cm	Chest cm
12.22	87.58	69.54	42.63	35.88	45.53	92.50
1.93	6.90	7.60	3.90	2.42	2.75	9.58
12.47	89.60	71.88	44.40	37.14	47.10	94.19
2.01	7.16	7.05	3.46	1.64	2.12	10.21
11.82	84.27	65.73	39.73	33.82	42.95	89.73
1.79	5.20	7.16	2.72	2.09	1.31	8.13

Append 3.2

Anthropometric Measurements on Occupants

Append 3.1

But - knee cm	But - sole cm	Floor - knee cm	Foot cm	Abdomen cm	Pelvic width cm	Pelvic circ cm
61	104	55	28	86	38	99
59	108	60	29	83	35	95
55	96	51	21	80	39	90
59	101	47	23	104	43	100
59	103	53	23	68	36	95
57	103	56	25	84	36	98.5
57	96	50	24	72	40	93
62	100	49	23	78	42	96
55	99	52	22	70	33	92
58	106.5	57.5	24	104	40	110
63	108	57	27	89	36	98
57	98	50	25	75	40	100
64	111	60	29	83	40	100
56	103	50	24	83	36	95
56	100	52	25	101	43	106
57	95	51	28	96	35	93
54	97	48	23	79	39	102
55	97	52	23	76	37	96
58	98	50	24	79	39	98
59	106	56.5	26.5	92	34.5	102
61	107	57	29	94	39	99
52	91	48	22	83	38	104
56	102	56	26	93	39	94
57.5	99	52	23.5	89	35	98
56	105	52.7	25	76.5	39.5	97
57.6	105	53.4	25	77	33	95.5
56	109	58	26	77	36	94
62	111	59	29	76	35	92
56	102	55	28	79	36	94

Average Anthropometric Measurements

But - knee cm	But - sole cm	Floor - knee cm	Foot cm	Abdomen cm	Pelvic width cm	Pelvic circ cm
57.76	102.09	53.38	25.17	83.67	37.66	97.45
2.81	5.07	3.76	2.38	9.67	2.73	4.46
58.45	104.69	55.45	26.50	86.81	37.00	97.78
2.65	4.20	3.04	1.93	8.53	2.57	4.62
56.64	97.82	50.00	23.00	78.55	38.73	96.91
2.80	3.12	1.90	1.10	9.57	2.76	4.35

Append 3.2

Table Append 3.3 compares anthropometric data for the hybrid III 50th percentile anthropomorphic test device with mean measurements obtained from occupants seated in the mid section of G-OBME. The differences between the occupants average anthropometric dimensions and that of the the 50th percentile Hybrid III dummy are expressed as percentages.

Table Append 3.3

Dimension	Hybrid III	Occupant	Difference %
Height	168.2	170	+1
Head breadth	15.7	15	-4.5
Sitting height	89.6	87.6	-2.2
Shoulder width	46.4	42.6	-10.4
Shoulder to olecranon	36.6	35.9	-1.4
Olecranon to fingertip	46.4	45 5	-1.9
Chest circumference	100.0	92.5	-7.5
Buttock/knee length	57.6	57.8	+0.3
Knee height (sitting)	56.4	53.4	-5.3
Foot length	26.0	25.2	-3
Abdominal circumference	79.0	83.7	+5.9

Dimensions expressed in cm.

As can be seen the average measurements of all occupants seated in the mid section of the aircraft compare favourably with the dimensions of a hybrid III 50th percentile dummy. This anthropomorphic test device is therefore a suitably sized model to investigate the kinematics for the occupants of G-OBME. However it must be remembered that the 50 percentile Hybrid III dummy used was a male surrogate.

Appendix 4

Calibration of Sliding Knee Potentiometers

Sliding knee potentiometers were placed in the the left and right knee assemblies of the hybrid III anthropomorphic test device. The potentiometers measure shear or tibial displacement in relation to the femoral condyles of the knee joint, as a result of impact to the region of the knee and tibial tuberosity. The potentiometers are used in the automobile industry to assess bolster impacts to the knee.

Method

Test facility and test fixture

The helmet lab at the Royal Air Force Institute of Aviation Medicine, Farnborough, Hampshire was used. This consists of a drop tower, from which a known weight can be dropped from a designated height, and a test bed.

The test fixture (figure Appen 4.1), designed for the purpose of calibration of the knee potentiometer by JM Rowles and L Neil, consisted of a bracket mounted to the test bed that held the femoral component of the thigh assembly of the hybrid III dummy. The design of the bracket allowed the assembly to be moved from a central position in order that the tibial tuberosity region of the tibia could be struck by an impactor.

Instrumentation and measurements

An accelerometer was mounted on the impactor to record the

Test Fixture for Calibration of Sliding Knee Potentiometers

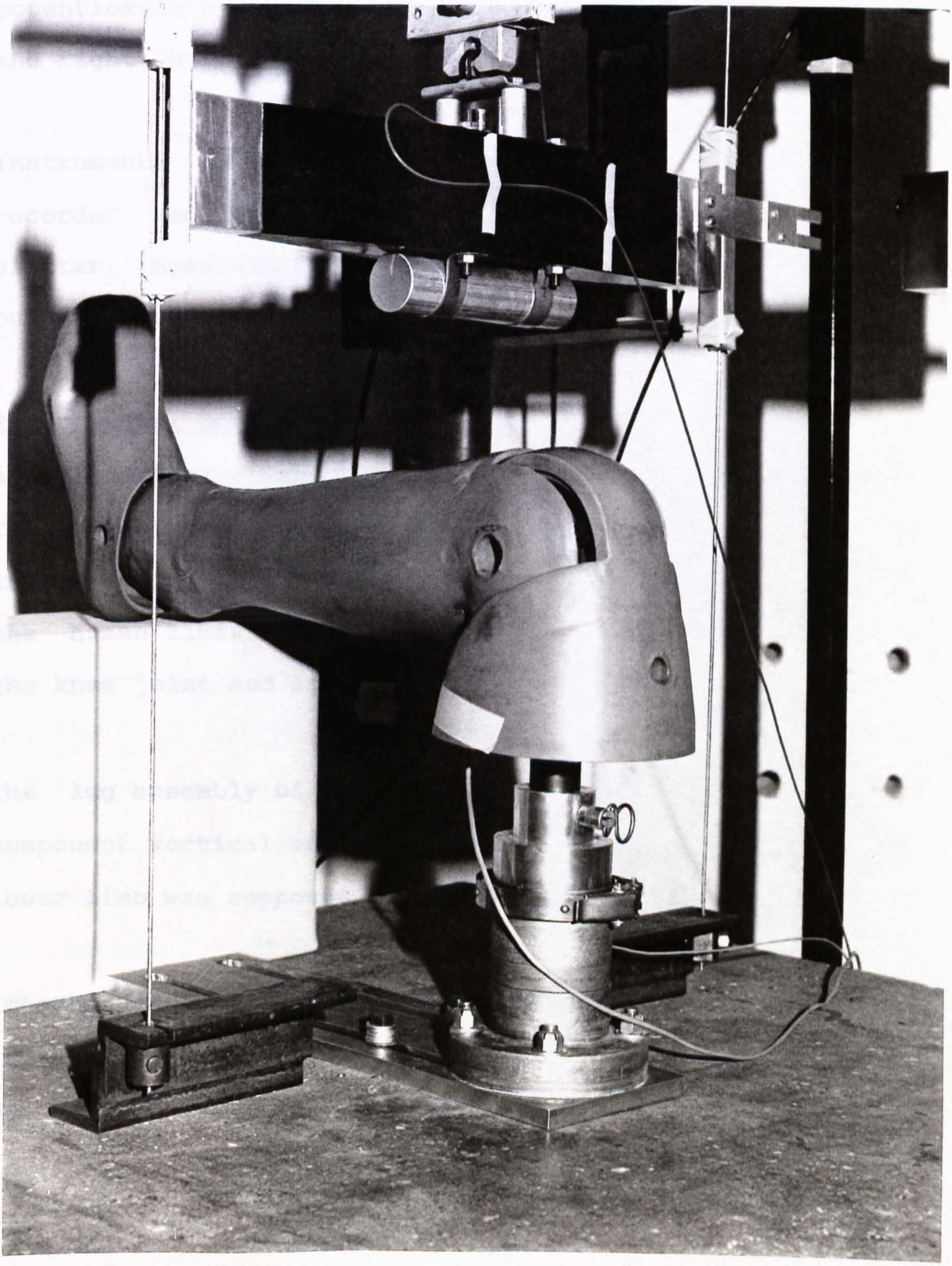

Figure Appen 4.1

acceleration on impact with the tibial tuberosity of the leg assembly. A load cell was placed in the femoral component of the limb in order to record the load transmitted through the knee to the femur. Sliding knee potentiometers were mounted in the knee assemblies of left and right lower limbs.

Instruments were connected to a Datalab 2000 Transient recorder and the data recorded to a Gouldes ES 1000 plotter. Measurements were taken directly from the the output of the recorder.

Experimental procedure

A cylindrical aluminium impactor of 1.5" radius was used. The weight of the impactor assembly was 6.041 kg. The impact point on the shin approximated to the position of the human tibial tuberosity and was 3" from the centre of the knee joint and 15" from the heel of the dummy.

The leg assembly of the dummy was placed with the femoral component vertical and the tibial component horizontal. The lower limb was supported at the heel by a foam block.

Impacts of various loads were achieved by dropping the impactor assembly from varying heights. These ranged from 8 cm to 1cm. At these heights readings from the knee potentiometers reflected those seen in the experimental impact simulations.

Each 'drop' height was repeated 5 times with all conditions randomised. Calibration was carried out on both right and left limbs.

Results

The means and standard deviations for each test condition are recorded in table Appen 4.2 for the right knee and Appen 4.3 for the left knee.

Figure Appen 4.4 and figure Appen 4.5 are calibration graphs for the right and left knee potentiometers. The line of best fit reveals a linear relationship between knee potentiometer readings and the load transmitted to the femoral shaft, and thus the load transmitted across the knee joint. Measuring the gradient of the line of best fit thus allows calibration of the knee potentiometers.

The left knee potentiometer unfortunately failed to function correctly and demonstrated a slow return or inability of the assembly to return to its resting position. As a result of this flaw the data recorded from the left knee potentiometer was unable to be manipulated by the computer soft ware developed to analyse the results. This was caused by an inability to identify the base line (resting state). Information from this knee potentiometer was therefore disregarded in the statistical analysis.

Appen 4.2

RIGHT KNEE SLIDING KNEE POTENTIOMETER

Drop ht (cm)	Femoral load (N) Mean	Std.	Acceler. (G) Mean	Std.	Knee pot (units) Mean	Std.
8.00	822.00	31.14	12.60	0.22	47.00	0.71
7.00	732.00	22.80	11.40	0.42	43.20	0.84
6.00	706.00	26.08	10.60	0.22	39.00	0.71
5.00	610.00	14.14	9.70	0.27	35.00	0.71
4.00	568.00	8.37	9.00	0.00	32.80	0.84
3.00	514.00	8.94	8.30	0.27	27.20	0.84
2.00	426.00	8.94	7.30	0.27	23.00	0.00
1.00	296.00	11.40	5.40	0.22	14.80	0.45

Appen 4.3

LEFT KNEE SLIDING KNEE POTENTIOMETER

Drop ht (cm)	Femoral load (N) Mean	Std.	Acceler. (G) Mean	Std.	Knee pot (units) Mean	Std.
8.00	834.00	26.08	13.10	0.42	47.80	1.64
7.00	802.00	34.21	12.40	0.22	44.80	1.30
6.00	752.00	45.50	12.20	0.27	42.00	1.22
5.00	726.00	35.07	11.50	0.00	38.00	1.00
4.00	676.00	26.08	10.80	0.27	35.00	0.71
3.00	592.00	8.37	10.10	0.22	29.60	1.14
2.00	500.00	12.25	8.70	0.27	23.60	0.89
1.00	390.00	12.25	6.90	0.22	17.00	0.71

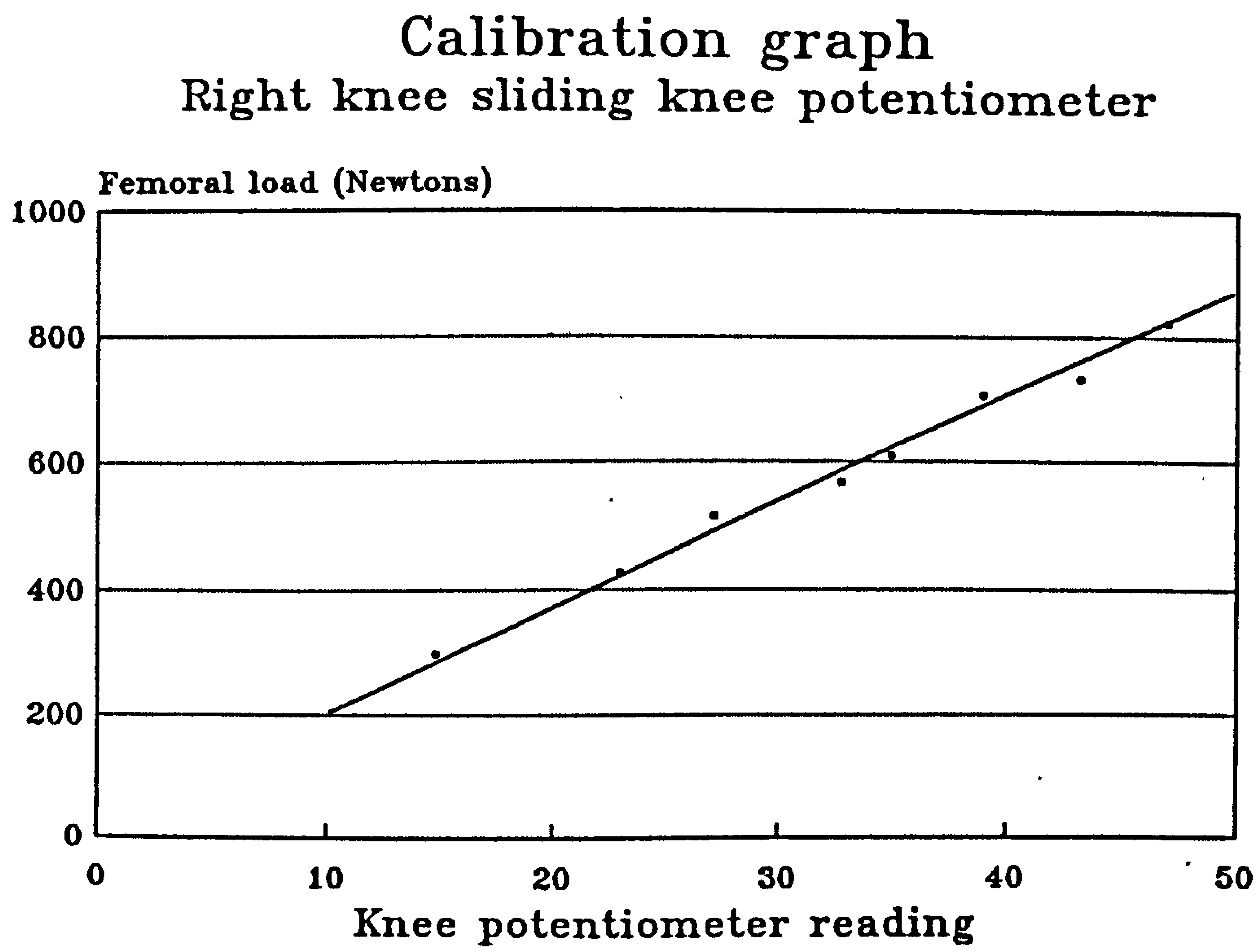

Figure Appen 4.4

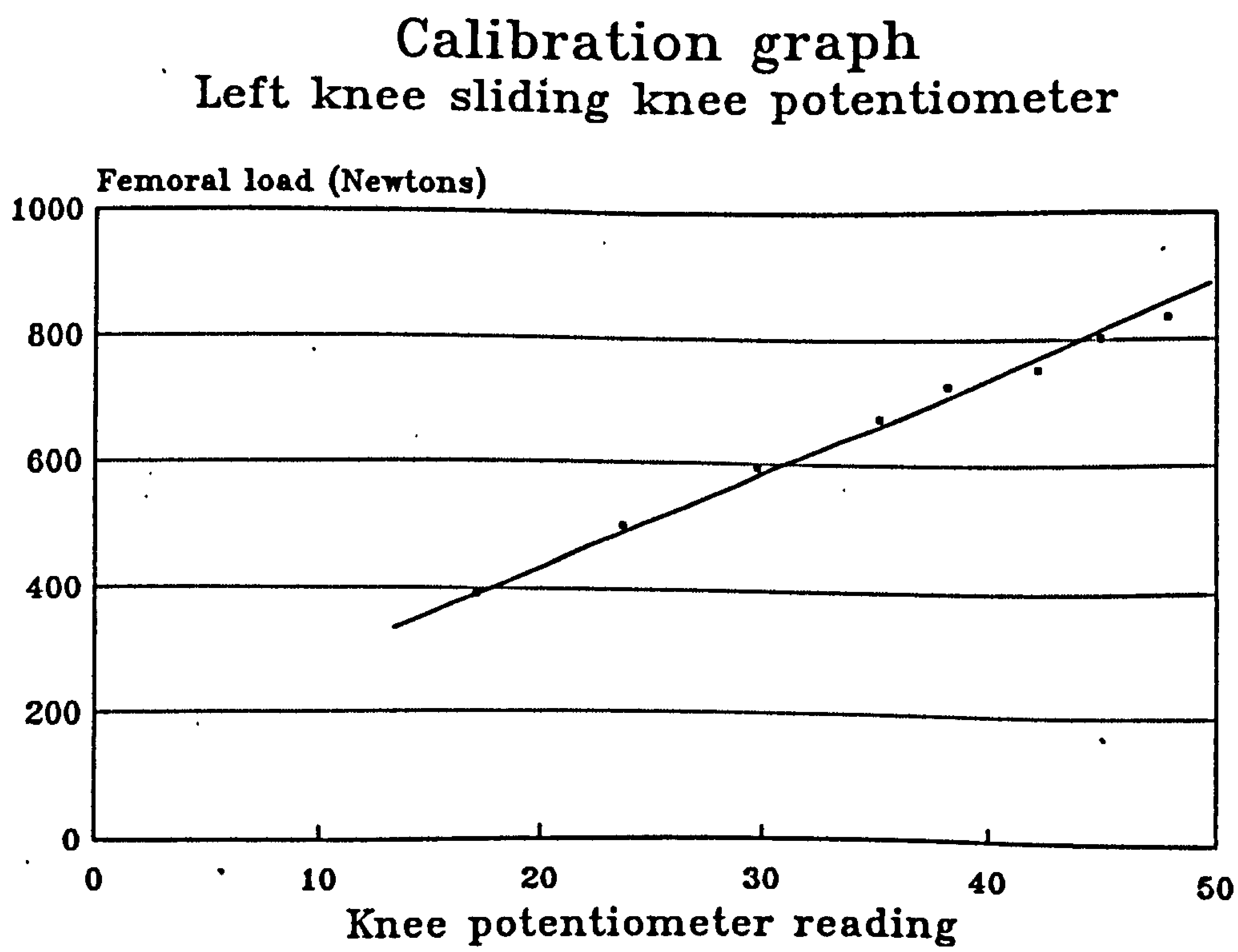

Figure Appen 4.5

Appendix 5

Sled Test Results

In total 135 experimental runs were undertaken, excluding calibration runs and runs lost as a result of equipment damage or failure. Eleven parameters in total were measured but a further three pelvic parameters were calculated from the results (table Appen 5.1). The results for each run are illustrated in table Appen 5.2.

Append 5.1

Parameter Measured	Abbreviation	Units
Vehicle impact pulse	V G	G
Pelvic Gx impact pulse	P Gx	G
Pelvic Gz impact pulse	P Gz	G
Lap belt load	STRap	kN
Left leg shear	LLeg	N
Right leg shear	RLeg	N
Pelvic X displacement	Pdis X	mm
Pelvic Z displacement	Pdis Z	mm
Thigh X displacement	Kdis X	mm
Thigh Z displacement	Kdis Z	mm
Ankle X displacement	Adis X	mm
Ankle Z displacement	Adis Z	mm

Calculated Pelvic Parameters

Maximum footwards acceleration	PGz pos	G
Maximum headwards acceleration	PGz neg	G
Resultant pelvic acceleration	PGres	G

RUN	TYPE	V G	P Gx	PGz POS	PGz NEG	PGz MAX	PGp RES	STRAP
		(G)	(G)	(G)	(G)	(G)	(G)	(kN)
3027	1111	8.85	9.81	3.33	1.11	3.33	9.91	2.76
3040	1111	8.88	10.28	3.33	2.04	3.33	10.37	2.59
3046	1111	8.81	9.16	2.87	2.41	2.87	9.26	2.65
3058	1111	9.08	9.72	3.33	2.69	3.33	9.91	2.76
3063	1111	8.85	10.00	2.69	3.06	3.06	9.91	2.81
3028	1112	8.92	10.37	3.80	1.67	3.80	10.46	2.89
3042	1112	8.85	8.97	3.43	2.31	3.43	9.26	2.39
3047	1112	8.88	9.63	3.98	2.41	3.98	9.72	2.43
3051	1112	9.15	9.44	2.87	2.87	2.87	9.44	2.58
3064	1112	8.81	9.44	4.44	2.50	4.44	10.00	2.41
3029	1121	9.00	10.84	4.63	2.22	4.63	10.93	2.91
3035	1121	9.35	9.44	3.43	2.22	3.43	9.44	2.42
3048	1121	8.88	9.53	3.89	2.96	3.89	9.72	2.53
3052	1121	9.23	10.09	3.80	2.50	3.80	10.19	2.59
3066	1121	8.92	10.47	3.24	1.94	3.24	10.37	2.73
3034	1122	9.12	9.81	4.17	2.31	4.17	10.28	2.61
3041	1122	8.85	9.16	4.44	2.04	4.44	9.54	2.18
3045	1122	9.04	9.53	4.17	2.59	4.17	9.81	2.43
3053	1122	8.81	9.16	3.52	1.85	3.52	9.26	2.52
3065	1122	9.27	9.81	3.70	2.31	3.70	9.91	2.73
3107	1123	9.00	9.25	2.78	2.13	2.78	9.17	2.48
3108	1123	9.04	10.47	3.24	2.41	3.24	10.37	2.66
3109	1123	8.65	10.75	3.80	2.59	3.80	10.74	2.76
3110	1123	9.00	11.31	3.61	2.50	3.61	11.30	2.86
3111	1123	9.04	10.75	3.24	3.33	3.33	10.65	2.85
3030	1211	9.08	8.79	1.85	3.06	3.06	8.80	2.99
3036	1211	9.00	8.50	1.85	3.33	3.33	8.80	3.01
3050	1211	8.96	8.79	2.41	4.26	4.26	9.35	3.33
3054	1211	9.00	7.76	2.31	3.80	3.80	7.87	3.02
3059	1211	9.23	7.85	3.33	3.61	3.61	8.15	2.99
3031	1212	9.58	9.91	2.13	3.89	3.89	10.00	3.75
3038	1212	8.77	7.66	2.69	3.70	3.70	7.69	2.90
3043	1212	8.65	8.22	1.76	2.22	2.22	8.33	2.82
3055	1212	8.81	8.50	2.22	3.61	3.61	8.43	3.21
3060	1212	8.85	7.57	1.94	3.24	3.24	7.69	3.30
3033	1221	9.04	9.07	1.57	5.09	5.09	10.00	3.72
3037	1221	9.42	7.85	1.85	2.96	2.96	8.06	2.76
3049	1221	8.96	8.60	0.65	4.44	4.44	9.07	3.22
3057	1221	8.92	8.88	2.22	3.33	3.33	9.26	3.36
3061	1221	8.96	8.50	2.96	2.69	2.96	8.70	3.23
3032	1222	9.42	9.72	2.13	5.28	5.28	9.81	3.84
3039	1222	8.88	8.69	2.41	2.96	2.96	8.89	3.02
3044	1222	8.77	8.69	1.30	2.31	2.31	8.61	2.91
3056	1222	8.85	7.94	2.50	3.33	3.33	8.06	3.14
3062	1222	8.92	9.25	1.67	3.15	3.15	9.17	3.28
3072	2111	14.31	19.25	7.41	3.98	7.41	19.54	5.10
3078	2111	15.46	18.13	7.31	3.52	7.31	18.24	4.97
3084	2111	14.92	19.16	6.30	5.74	6.30	19.07	5.00
3097	2111	15.12	19.44	8.70	3.61	8.70	19.72	5.08
3105	2111	14.81	18.41	6.76	4.26	6.76	18.43	5.00
3071	2112	14.58	18.13	7.69	4.35	7.69	18.15	4.85
3077	2112	14.92	17.76	7.22	4.81	7.22	17.87	5.05
3083	2112	15.12	18.60	9.07	4.26	9.07	19.35	4.82
3094	2112	14.92	19.07	8.80	3.33	8.80	19.26	5.00
3102	2112	15.12	18.97	7.50	4.72	7.50	18.98	4.90
3070	2121	14.88	18.88	5.93	6.11	6.11	18.89	4.92
3076	2121	14.88	18.69	8.80	5.19	8.80	18.80	4.70
3089	2121	14.54	19.16	6.39	6.11	6.39	19.54	4.85
3098	2121	14.85	19.35	7.78	6.20	7.78	19.54	5.10
3101	2121	15.04	16.45	4.54	3.52	4.54	16.48	4.82
3073	2122	14.69	17.48	8.52	4.81	8.52	17.96	4.55
3082	2122	14.77	18.97	8.24	4.07	8.24	19.17	4.95
3090	2122	14.88	19.35	8.70	5.93	8.70	19.17	5.23
3092	2122	14.92	18.79	8.43	4.63	8.43	18.70	5.00
3103	2122	15.12	18.60	7.22	4.63	7.22	18.52	4.77
3112	2123	15.27	19.07	7.22	5.65	7.22	19.44	5.03

3113	2123	14.54	19.16	5.74	6.30	6.30	18.98	5.25
3114	2123	15.08	18.88	7.96	4.81	7.96	19.07	5.00
3115	2123	14.88	19.72	8.80	6.57	8.80	19.91	5.23
3116	2123	15.04	19.53	6.67	6.67	6.67	19.54	5.23
3069	2211	14.85	16.26	3.06	6.02	6.02	16.11	6.01
3075	2211	15.12	15.89	2.87	6.02	6.02	15.74	5.88
3087	2211	15.08	15.51	3.43	5.83	5.83	15.56	5.96
3095	2211	15.12	16.07	3.15	4.81	4.81	16.02	5.68
3106	2211	14.73	16.36	2.87	5.00	5.00	16.30	6.14
3068	2212	14.96	16.82	3.15	6.02	6.02	16.85	6.09
3080	2212	15.46	15.79	3.80	5.93	5.93	16.30	6.24
3086	2212	14.85	15.70	3.33	6.11	6.11	15.65	6.04
3093	2212	15.12	16.07	3.61	5.93	5.93	16.20	6.19
3099	2212	14.92	16.54	3.06	6.57	6.57	16.48	6.21
3074	2221	14.65	16.36	2.50	5.00	5.00	16.30	5.86
3081	2221	15.00	15.61	3.24	5.74	5.74	15.65	5.53
3088	2221	15.27	17.20	3.33	4.72	4.72	17.04	5.45
3091	2221	14.23	15.79	1.85	4.54	4.54	15.74	5.48
3100	2221	14.96	14.86	2.87	5.19	5.19	14.91	5.56
3067	2222	15.42	16.73	2.96	5.56	5.56	16.57	5.78
3079	2222	14.96	16.17	2.87	5.37	5.37	16.11	5.56
3085	2222	15.19	14.86	2.59	5.74	5.74	14.91	5.45
3096	2222	15.19	16.54	3.43	5.19	5.19	16.39	5.78
3104	2222	14.62	16.54	2.96	4.63	4.63	16.39	5.51
3121	3111	20.31	26.45	11.30	5.00	11.30	27.04	7.58
3125	3111	20.38	26.82	8.80	4.54	8.80	27.13	7.45
3138	3111	20.31	25.98	10.37	5.37	10.37	26.20	7.98
3152	3111	20.31	27.01	9.17	5.09	9.17	27.22	8.08
3157	3111	19.85	26.26	10.46	5.83	10.46	26.67	7.75
3120	3112	20.23	26.17	10.65	4.44	10.65	26.67	7.60
3132	3112	20.15	25.61	9.54	3.89	9.54	26.11	7.75
3137	3112	20.08	25.89	10.09	4.26	10.09	26.20	7.93
3148	3112	20.38	26.64	10.74	5.09	10.74	26.67	8.08
3159	3112	20.46	26.36	10.74	4.72	10.74	26.48	7.50
3118	3121	19.00	26.54	13.06	7.87	13.06	27.41	7.50
3131	3121	20.00	27.20	6.76	5.37	7.22	27.22	8.16
3135	3121	20.08	26.45	7.87	7.22	7.87	26.67	7.95
3151	3121	20.00	26.36	11.67	5.56	11.67	27.22	7.45
3160	3121	20.62	27.29	11.30	7.59	11.30	28.06	7.58
3119	3122	19.46	25.98	9.81	5.00	9.81	26.85	7.55
3130	3122	19.77	26.54	9.72	6.94	9.72	27.59	7.60
3143	3122	20.46	26.45	10.37	4.81	10.37	27.04	7.75
3150	3122	19.85	26.45	8.70	6.20	8.70	26.67	7.78
3158	3122	20.31	26.07	10.93	4.63	10.93	27.04	7.42
3133	3123	20.00	27.01	6.85	5.74	6.85	27.04	8.23
3134	3123	20.23	26.82	8.15	6.76	8.15	27.04	8.08
3139	3123	20.00	27.57	12.31	6.20	12.31	28.24	7.63
3147	3123	19.92	26.54	7.96	3.15	7.96	26.76	8.31
3156	3123	20.38	25.89	8.33	4.44	8.33	26.11	7.85
3117	3211	18.85	23.55	4.63	7.13	7.13	23.33	8.48
3129	3211	19.77	23.55	4.26	7.69	7.69	23.33	8.79
3141	3211	19.38	23.74	2.96	6.76	6.76	23.52	9.19
3145	3211	20.08	24.67	4.44	7.41	7.41	24.54	9.22
3161	3211	20.23	23.55	2.96	7.69	7.69	23.33	8.89
3124	3212	20.15	23.55	4.17	6.85	6.85	23.33	8.79
3128	3212	20.46	22.80	4.81	7.22	7.22	22.78	9.12
3142	3212	20.46	24.49	4.91	7.87	7.87	24.26	8.76
3149	3212	19.92	22.62	5.28	7.69	7.69	22.78	8.61
3153	3212	20.38	24.39	4.17	8.24	8.24	24.17	8.69
3122	3221	20.15	25.42	2.78	5.28	5.28	25.19	8.36
3127	3221	19.85	23.93	2.87	5.09	5.09	23.70	8.33
3136	3221	20.15	23.27	3.52	7.04	7.04	23.06	8.46
3146	3221	20.38	22.71	3.61	6.76	6.76	22.50	8.84
3155	3221	19.85	25.33	5.00	6.30	6.30	25.09	8.06
3123	3222	19.92	24.67	4.17	6.30	6.30	24.44	8.51
3126	3222	19.77	23.93	4.44	6.67	6.67	23.70	8.33
3140	3222	20.46	24.39	4.91	6.20	6.20	24.17	8.06
3144	3222	20.38	24.39	3.43	6.85	6.85	24.17	8.36
3154	3222	19.92	24.58	4.63	6.20	6.20	24.35	8.43

TYPE	L LEG (Units)	R LEG (Units)	Pdis X (mm)	Pdis Z (mm)	Kdis X (mm)	Kdis Z (mm)	Adis X (mm)	Adis Z (mm)
1111	2	7	170	10	145	11	145	15
1111	3	4	195	9	147	9	165	14
1111	2	8	165	11	117	11	135	13
1111	2	11	175	12	135	10	120	15
1111	2	18	170	11	132	7	163	6
1112	2	9	169	8	130	10	143	14
1112	2	12	160	17	105	7	125	12
1112	2	15	160	9	110	12	200	11
1112	3	5	188	17	140	13	175	13
1112	2	10	170	8	118	10	175	12
1121	2	17	180	18	130	50	350	110
1121	2	7	210	14	145	70	315	45
1121	2	22	185	17	125	53	315	95
1121	2	31	207	14	150	60	355	80
1121	2	29	190	15	130	55	330	85
1122	2	9	140	14	97	35	310	110
1122	2	26	170	13	109	50	330	95
1122	3	31	165	12	108	42	330	105
1122	2	15	185	11	125	46	320	90
1122	2	23	165	12	125	45	330	100
1123	8	3	192	19	155	15	149	12
1123	17	3	200	22	165	21	178	17
1123	14	3	198	25	162	36	220	15
1123	13	4	194	23	158	59	260	25
1123	34	33	205	20	170	52	240	21
1211	2	9	153	14	132	21	98	9
1211	2	10	180	29	152	22	98	7
1211	3	6	175	24	142	25	77	9
1211	2	8	155	14	123	19	98	10
1211	2	4	190	20	150	21	95	11
1212	2	8	145	15	120	15	82	5
1212	2	11	155	29	130	20	67	4
1212	2	6	147	18	115	15	89	7
1212	2	8	145	24	110	15	85	6
1212	2	7	140	20	117	14	83	5
1221	2	16	160	17	140	10	113	8
1221	2	6	170	16	140	10	120	10
1221	2	13	165	20	145	8	102	8
1221	2	13	155	18	125	11	95	4
1221	2	4	185	21	150	20	160	9
1222	2	4	133	16	113	10	85	10
1222	2	3	145	13	120	42	230	31
1222	3	10	145	20	122	36	290	95
1222	3	10	136	20	105	16	100	10
1222	2	12	142	17	109	40	268	85
2111	18	9	201	29	160	20	160	45
2111	9	11	190	21	150	11	166	14
2111	14	4	200	25	165	9	198	16
2111	20	7	200	27	155	12	171	16
2111	11	5	203	35	162	13	154	18
2112	18	13	175	20	135	16	215	19
2112	19	4	180	22	144	10	175	15
2112	19	6	180	21	139	13	213	23
2112	13	6	180	27	145	10	180	16
2112	14	5	195	29	151	17	171	25
2121	3	86	228	29	175	100	360	92
2121	5	29	222	24	158	93	343	90
2121	2	86	235	32	178	100	310	75
2121	3	55	212	31	168	95	345	77
2121	9	7	212	39	162	61	211	20
2122	4	35	198	20	140	74	335	115
2122	11	37	195	22	143	86	360	85
2122	2	63	197	32	147	90	305	98
2122	13	41	196	26	142	81	332	104
2122	12	14	204	30	157	95	330	85
2123	14	27	226	32	155	110	365	65

2123	11	76	220	28	184	115	375	98
2123	3	100	237	38	184	112	383	105
2123	32	52	233	35	186	112	365	110
2123	10	102	220	33	182	115	350	65
2211	4	10	199	25	160	26	130	20
2211	6	10	202	30	168	26	125	18
2211	17	7	217	40	182	31	136	20
2211	12	5	185	38	150	24	133	23
2211	15	7	175	40	138	26	104	14
2212	2	9	170	31	130	25	130	23
2212	5	12	199	27	160	25	123	17
2212	2	51	190	29	156	22	138	16
2212	33	6	182	38	144	25	113	13
2212	23	7	178	36	136	25	97	18
2221	19	4	178	32	144	41	220	16
2221	7	13	205	33	170	31	162	18
2221	14	17	183	42	147	75	250	15
2221	31	5	182	38	145	35	168	19
2221	14	12	221	38	182	17	143	12
2222	4	3	198	33	152	40	180	21
2222	1	1	175	28	141	57	230	15
2222	2	4	199	32	160	31	175	17
2222	16	80	169	38	138	87	268	22
2222	14	6	170	39	147	28	122	17
3111	35	4	227	34	185	68	366	35
3111	19	5	239	38	199	59	330	25
3111	28	4	215	34	185	65	218	28
3111	31	4	227	35	193	18	258	23
3111	42	6	210	39	170	16	225	22
3112	35	3	215	36	172	12	253	23
3112	37	5	213	31	179	7	220	19
3112	25	7	215	35	172	18	209	20
3112	26	5	213	40	169	11	215	25
3112	38	4	220	38	174	11	218	28
3121	14	60	226	41	202	120	380	97
3121	2	58	224	37	186	125	347	66
3121	47	69	230	34	197	130	348	57
3121	7	44	231	36	182	131	378	90
3121	28	78	222	37	179	125	375	113
3122	15	47	212	33	167	109	350	91
3122	9	45	209	35	163	118	355	64
3122	24	41	209	39	171	110	366	82
3122	4	51	229	38	193	122	384	82
3122	33	69	223	37	180	116	378	87
3123	9	69	238	37	196	125	356	62
3123	6	43	228	33	190	126	346	67
3123	30	71	228	36	190	109	375	118
3123	25	120	220	35	185	118	355	90
3123	12	154	233	39	195	127	335	67
3211	23	9	192	40	160	21	161	17
3211	21	4	208	35	175	21	263	14
3211	24	15	208	40	179	25	158	18
3211	24	10	205	44	179	21	160	16
3211	41	16	220	45	181	27	152	20
3212	14	9	215	38	173	23	154	15
3212	23	16	221	49	189	29	152	17
3212	28	12	201	40	169	25	162	18
3212	22	11	216	46	181	24	165	17
3212	28	13	199	44	165	23	154	18
3221	26	161	227	36	195	110	319	62
3221	7	198	220	47	180	130	320	66
3221	40	12	229	41	200	64	251	19
3221	31	11	207	45	182	56	230	21
3221	29	10	220	40	190	102	294	21
3222	10	169	200	41	165	115	300	68
3222	21	4	203	44	170	56	233	20
3222	56	212	210	43	175	112	325	64
3222	48	210	199	42	152	114	310	70
3222	5	171	196	48	163	121	305	59